# Rumpfmuskelkoordination: Reaktion auf dynamische und statische externe Stimuli

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## Abkürzungsverzeichnis

| | |
|---|---|
| CURS | Chronisch unspezifischer Rückenschmerz |
| EMG | Elektromyographie |
| ES | Musculus erector spinae |
| F/B | engl.: front/back<br>Verhältnis der ventralen zu den dorsalen Rumpfmuskeln |
| ME | Motorische Einheit |
| MF | Musculus multifidus |
| MVC | engl.: maximum voluntary contraction<br>Isometrische Maximalkraft |
| NEMG | Nadel-Elektromyographie |
| OE | Musculus obliquus externus abdominis |
| OEMG | Oberflächen-Elektromyographie |
| OI | Musculus obliquus internus abdominis |
| RA | Musculus rectus abdominis |
| RMS | engl.: root mean square<br>Wurzel des Mittelwertes der Amplitudenquadrate |
| TA | Musculus transversus abdominis |
| WS | Wirbelsäule |

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# 1 Einführung

Die Wirbelsäule (WS) bildet das Achsenskelett des menschlichen Rumpfes (Voss und Herrlinger 1979, Rauber und Kopsch 2003). Ihr kommt somit eine wesentliche Stützfunktion zu. Gleichzeitig ermöglicht sie aufgrund ihres Aufbaues aus insgesamt 24 Einzelwirbeln eine erhebliche Gesamtbeweglichkeit, die Bewegungen des Rumpfes in der Sagittal-, der Frontal- und der Horizontalebene, also allen Hauptbewegungsebenen ermöglicht. Sie liegt dorsal im Rumpfquerschnitt und weist in der Sagittalebene eine doppelt s-förmige Form auf.

An der Aufrechterhaltung des dynamischen Gleichgewichtes der Wirbelsäule zwischen Bewegung und Stabilität sind verschiedene Komponenten beteiligt. Dies sind in grober Einteilung passive und aktive Strukturen (Panjabi 1992b, Panjabi 1992a). Die passiven Strukturen gliedern sich weiterhin in knöcherne und ligamentäre Bestandteile auf.

Die anhand der Bewegungen des Oberkörpers erkennbare Gesamtbeweglichkeit der Wirbelsäule setzt sich aus den Einzelbewegungen der Wirbel gegeneinander zusammen. Die funktionelle Einheit zweier benachbarter Wirbelkörper mit ihren Gelenken, den anliegenden Abschnitten der ligamentären Strukturen, der Bandscheibe und den umgebenden Muskeln samt ihrer Innervation wird als Bewegungssegment definiert (Voss und Herrlinger 1979, Rauber und Kopsch 2003).

Die Wirbelkörper artikulieren direkt über die Intervertebralgelenke miteinander. Diese weisen in den verschiedenen Abschnitten der WS eine unterschiedliche Stellung auf und ermöglichen so bevorzugte Bewegungsrichtungen: im zervikalen Bereich sind alle drei Bewegungsebenen möglich, thorakal kann bevorzugt rotiert werden, während lumbal vor allem die Bewegungen in der Frontal- und Sagittalebene möglich sind (Voss und Herrlinger 1979, Rauber und Kopsch 2003). Das mögliche Ausmaß von Bewegungen gegeneinander variiert ebenfalls regional. Beispielsweise sind im thorakalen Bereich Rotationswinkel von bis zu 102° zwischen benachbarten Wirbeln möglich (Gregersen und Lucas 1967). Bewegungen in der Frontalebene sind praktisch immer mit rotatorischen Bewegungen verbunden (Voss und Herrlinger 1979, Rauber und Kopsch 2003).

Alle Bewegungen werden durch die wasserkissenartige Verbindung zwischen benachbarten Wirbelkörpern, der Bandscheibe, bestehend aus dem Nucleus pulposus und dem Anulus fibrosus ermöglicht.
Die ligamentären Bestandteile der passiven Strukturen sichern die Integrität der Wirbelsäule vor allem in den Grenzbereichen der möglichen Bewegungen.
Die aktive Struktur zur Sicherung der Wirbelsäulenintegrität wird von der die WS umspannenden Muskulatur gebildet. Sie ermöglicht durch ihre fein koordinativ abgestimmte Aktivierung die erforderliche Stabilität, solange der Bewegungsbereich nicht ausgeschöpft wird. Dies betrifft die übergroße Mehrheit alltäglicher Aktivitäten. Gleichzeitig wird durch die Muskelwirkung erst jegliche aktive Bewegung der Wirbelsäule ermöglicht. Dabei haben die Muskeln, je nach ihrer Anordnung mehr stabilitätsvermittelnde oder bewegungsvermittelnde Aufgaben.

## 1.1 Systematik der Rumpfmuskeln

Anhand der anatomischen Literatur (Valerius et al. 2002, Rauber und Kopsch 2003, Voss und Herrlinger 1979) kann man sich sehr leicht einen Überblick über die entsprechend der Ansatz- und Ursprungsorte rein biomechanisch herzuleitende Funktion der Rumpfmuskeln verschaffen. Dort sind auch bereits funktionsbezogene Angaben zu finden, die die anatomischen Angaben ergänzen. Als Beispiel sei hier die Funktion der beiden schrägen Bauchmuskeln, der beiden Mm. obliqui abdominales internus et externus genannt: hier wirken der gleichseitige M. obliquus externus und der gegenseitige M. obliquus internus synergistisch bei der Drehung der Schulter nach vorn (Rauber und Kopsch 2003, Voss und Herrlinger 1979). Die klassisch-anatomische Einteilung der Rumpfmuskeln erfolgt anhand ihrer Lokalisation in Bauch- und Rückenmuskeln, sowie insbesondere für die Rückenmuskeln anhand ihrer Innervation durch die ventralen (oberflächliche Muskeln) oder dorsalen (tiefe, autochthone Muskeln) Äste der Spinalnerven. Insbesondere die autochthone Rückenmuskulatur wird weiterhin in Gruppen unterteilt, die durch die Ansatz- und Ursprungsorte definiert sind. Es werden folgende Gruppen unterschieden: die interspinale Gruppe, die spinale Gruppe, die transversospinale Gruppe, die intertransversale Gruppe und die spinotransversale Gruppe. Diese nachweisbare Vielfalt verschiedenster

Muskeln weist jedoch eine weitgehende funktionelle Redundanz auf, die weniger im Rahmen initiierter Bewegungen, sondern vielmehr für die Sicherung der segmentalen Stabilität zu suchen ist. Insofern bietet sich eine mehr integrative, und somit allgemeinere Unterteilung der Rumpfmuskeln an. Diese soll im Folgenden beschrieben werden.

Die Einteilung der die WS umspannenden Muskulatur zu so genannten Muskelsystemen geht maßgeblich auf die Ergebnisse von A. Bergmark im Jahr 1989 zurück (Bergmark 1989). Er unterteilte die Muskeln in die folgenden zwei Gruppen: das lokale und das globale Muskelsystem.

Dabei sind die Muskeln des lokalen Systems dadurch gekennzeichnet, dass sich sowohl ihr Ansatz als auch ihr Ursprung an der Wirbelsäule selber befinden. Als Vertreter dieser Gruppe sind demnach beispielsweise die genannten autochthonen Rückenmuskeln zu nennen. Ihre Aufgabe besteht definitionsgemäß in der Sicherung der Integrität des Bewegungssegmentes. Die dazu notwendigen Kräfte werden durch die erreichte "stiffness", also die Resistenz gegen exzentrische Belastung, bereits während geringgradiger Aktivierungsniveaus erreicht (Hoffer und Andreassen 1981). Eine Initiierung von Bewegungen kann bei der lediglich geringen Muskelquerschnittsfläche aufgrund des kurzen Hebelarmes durch diese Muskeln nicht geleistet werden. Sie sind durch ihren hohen oxidativen Faseranteil (Typ 1 Fasern) für tonische Aktivierungen auf vergleichsweise niedrigem Kraftniveau prädestiniert (Schilling 2005, Jorgensen et al. 1993, McFadden et al. 1984). Innerhalb dieses Systems sind sowohl mono- als auch polysegmentale Muskeln berücksichtigt, das heißt, sowohl lediglich ein Segment als auch mehrere Segmente überspannende Muskeln.

Die globalen Muskeln verbinden den Thorax mit dem Becken und vermitteln die Mobilität des Oberkörpers. Diese Muskeln sind aufgrund der anatomischen Anordnung und ihrer Faserzusammensetzung mit hohen Anteilen glykolytischer Fasern (Typ 2 Fasern) in der Lage hohe Kräfte zu generieren, sind jedoch stärker ermüdbar.

Da die Einteilung von Bergmark primär biomechanisch determiniert war, wurde in der Folge eine stärker funktionell orientierte Einteilung mit einer weiteren Aufteilung des globalen Systems in die global stabilisierenden und die global mobilisierenden Muskeln vorgenommen (Comerford und Mottram 2001). Die Funktionsspezifik der verschiedenen Rumpfmuskeln wird innerhalb des nunmehr erweiterten Systems der Rumpfmuskeln vor allem danach vorgenommen, inwiefern eine tonische bzw. phasische Aktivitätscharakteristik zu verzeichnen ist, das heißt, ob identifizierbare Aktivitäten der Muskulatur kontinuierlich und unabhängig von durchgeführten Bewegungen oder diskontinuierlich, nur im Zusammenhang mit Bewegungen vorhanden sind. Weiterhin spielt insbesondere für die Beurteilung der diskontinuierlichen Aktivitäten die Kontraktionsart im Sinne exzentrischer oder konzentrischer Kontraktionen der entsprechenden Muskeln eine wesentliche Rolle. Tabelle 1.1 gibt einen Überblick über die Zuordnungssystematik und weist wesentliche Muskeln den jeweiligen Systemen zu.

**Tabelle 1.1 Systematik der Rumpfmuskeln anhand funktioneller Charakteristika nach Comerford und Mottram (Comerford und Mottram 2001)**

| Name | Lokale Muskeln | global stabilisierende Muskeln | global mobilisierende Muskeln |
|---|---|---|---|
| Funktions-charakteristik | hohe Steifheit für Kontrolle der Segmentbewegungen<br><br>geringfügige bis keine Längenänderung bei Kontraktion<br>Aktivität unabhängig von Bewegung(-srichtung) | Kontrolle des Bewegungsausmaßes<br><br>Kontraktionsform: exzentrisch<br><br>Aktivität abhängig von Bewegungsrichtung | Kraftgenerierung für Erreichen des Bewegungsausmaßes<br>Kontraktionsform: konzentrisch<br><br>Aktivität abhängig von Bewegungsrichtung |
| zeitliche Aktivierung | kontinuierliche Aktivität | diskontinuierliche Aktivität | diskontinuierliche Aktivität |
| Beispiele | M. transversus abdominis<br>tiefe Anteile des M. multifidus (lumbalis) | Mm. obliqui abdominis<br>M. glutaeus medius<br>M. spinalis | M. rectus abdominis<br>M. erector spinae:<br>M. iliocostalis |

Die vorgestellte Systematik ist allgemein anerkannt und wird in wesentlichen Arbeiten zur Physiologie und Pathophysiologie der Rumpfmuskelfunktion verwendet (Wilke et al. 2003, Danneels et al. 2001, Solomonow et al. 2003, Keller et al. 2007, Lee et al. 2006, Hodges und Richardson 1996)

## 1.2 Klinischer Bezug

Die Einteilung der Rumpfmuskeln anhand der vorgestellten Systeme bildet die physiologische Grundlage für moderne physiotherapeutische Behandlungsstrategien des chronisch unspezifischen Rückenschmerzes (CURS) (Richardson et al. 1999).

Obwohl bei einer Lebensprävalenz von Rückenschmerzen zwischen 50-80% (Walsh et al. 1992, Brown et al. 1998) davon ausgegangen werden kann, dass fast jeder Mensch mindestens einmal in seinem Leben über Rückenschmerzen klagt, hat diese Erkrankung, wenn sie nicht durch spezifische Ursachen bedingt ist, immerhin eine Spontanheilungsrate von 25 % bis 58 % (Hestbaek et al. 2003), bezogen auf das Vorhandensein von Schmerzen nach einem Jahr. Eine weitere Krankschreibung 6 Monate nach dem akuten Schmerzereignis betrifft lediglich einen Anteil zwischen 3 % und 40 % der Patienten (Hestbaek et al. 2003). Trotz der generell guten Prognose akut auftretender Rückenschmerzen werden pro Jahr in den industrialisierten Ländern zweistellige Milliardenbeträge, vor allem durch die Zahl der verbleibenden Patienten mit chronifizierten Rückenschmerzen verursacht (Bolten et al. 1998, Göbel 2001, Ekman et al. 2005). Diese Patienten weisen keine systematisch mit dem Krankheitsbild einher gehenden pathologisch-anatomischen Veränderungen der WS auf (Phillips et al. 1986, van Tulder et al. 1997). Die großen Streuungen in den genannten Häufigkeitsangaben verweisen jedoch auf ein grundsätzliches Problem in der Definition des Krankheitsbildes (Müller 2001). Neben der das Krankheitsbild bestimmenden Schmerzsymptomatik weisen sie vor allem funktionelle Symptome auf. Dies betrifft psychische Variablen und Kennwerte der motorischen Leistungsfähigkeit.

Die Abweichungen der psychischen Variablen spiegelt sich vor allem in der Ausprägung von Depressivität und Ängstlichkeit (Rush et al. 2000, Polatin et al. 1993), sowie verstärktem Vermeidungsverhalten (Pfingsten und Schops 2004, Al-Obaidi et al. 2005) wieder. Ebenso existiert auch ein sehr deutlicher Zusammenhang zu Arbeitsunzufriedenheit, allgemeinen Arbeitsproblemen, mangelnder Unterstützung seitens der Vorgesetzten und anderen (Krause et al. 1998). Diese Störungsebenen sind nicht Thema der Arbeit und sollen deshalb nur der Vollständigkeit halber an dieser Stelle genannt sein.

Die motorische Leistungsfähigkeit setzt sich aus folgenden Komponenten zusammen: Kraft, Schnelligkeit, Ausdauer, Dehnbarkeit und Koordination (Hollmann et al. 2000). Die internationale Literatur konnte in großer Klarheit Nachweise erbringen, dass insbesondere die Kraft (Kraftausdauer) vermindert (Kankaanpää et al. 1998, da Silva et al. 2005, Saur et al. 1997) und die Koordination der untersuchten Rumpfmuskeln bei CURS-Patienten gestört sind (Anders et al. 2005, Hodges und Richardson 1999, Hodges und Richardson 1998).

Die Defizite im Kraft und Kraftausdauerbereich stehen im Zusammenhang mit der Tatsache, dass die Erkrankungshäufigkeit in den industrialisierten Ländern im Vergleich zu industriell weniger entwickelten Staaten deutlich erhöht ist (Volinn 1997). Dies kann am ehesten als Ergebnis dauerhafter Dekonditionierung weiter Teile der Bevölkerung angesehen werden (Volinn 1997). Andererseits korrelieren körperlich extrem belastende oder in nicht veränderbarer Zwangshaltung auszuübende Tätigkeiten (Van Nieuwenhuyse et al. 2006, Hartvigsen et al. 2001) mit dem Auftreten von Rückenschmerzen.

Die nachweisbaren Störungen der Rumpfmuskelkoordination betreffen vor allem die Muskeln des lokalen Systems und hier vor allem die Aktivierung des M. transversus abdominis (TA). Es konnten insbesondere verzögerte Aktivierungen, sowohl während willentlich initiierter Bewegungen (Hodges und Richardson 1998), als auch während extern reflektorisch provozierter Aktivierungen identifiziert werden (Hodges et al. 2001).

Diese Ergebnisse bilden die Grundlage für den bisher erfolgreichsten physiotherapeutischen Behandlungsansatz des CURS – das "spinal segmental stabilization program" (Richardson et al. 1999). Dabei werden unter Anleitung insbesondere die Muskeln des lokalen Systems zunächst willentlich aktiviert. Im weiteren Verlauf der Therapie wird dann versucht, diese willkürliche Ansteuerung zunehmend zu automatisieren und in Alltagsaktivitäten zu überführen. Obwohl bereits bei der Durchführung einer einfachen Rumpfhebe aus der Rückenlage alle Bauchmuskeln erheblich angespannt werden konnte eine aktuelle Arbeit nachweisen, dass die beobachtbaren Latenzzeiten für die Aktivierung des TA konsistent nur durch ein isoliertes Training dieses Muskels verkürzt werden konnten (Tsao und Hodges 2007).

Da sich chronische Rückenschmerzen jedoch praktisch immer aus einem akuten Rückenschmerzgeschehen entwickeln (Pfingsten und Schops 2004), kann die Frage bisher nicht eindeutig beantwortet werden, ob sowohl die beobachtbaren Defizite in der Kraftausdauerleistungsfähigkeit der CURS-Patienten als auch die identifizierten Koordinationsstörungen Ursache oder Ergebnis des Schmerzgeschehens sind.

Überlegungen anderer Autoren (Panjabi 2002, Panjabi 2003) gehen davon aus, dass als Ergebnis veränderter motorischer Kontrolle Veränderungen des normalen Bewegungsverhaltens folgen, die zu Mikrotraumen der WS führen. Die dadurch entstandenen Schmerzen wirken auf die Organisation der motorischen Kontrolle zurück (van Dieen et al. 2003a) und perpetuieren somit die vorbestehende oder dadurch ausgelöste Koordinationsstörung.

Es gibt jedoch Hinweise aus eigenen Untersuchungen (Anders et al. 1998), dass Personen ohne Rückenschmerzen bereits vor dem initialen Schmerzereignis Störungen der motorischen Kontrolle aufweisen können. Probanden, die während eines einmaligen dynamischen Kraftausdauertests der Rückenstrecker eine extrem starke muskuläre Ermüdung aufwiesen, veränderten dabei das lokale Aktivierungsmuster der Rückenmuskeln für die untersuchten Wirbelsäulensegmente, während Personen mit eher geringer Muskelermüdung das ursprüngliche Aktivierungsmuster bis zum Ende der Belastung beibehielten (Abbildung 1.1). Somit weist insbesondere die segmentbezogene Koordination der Rückenmuskulatur eine Abhängigkeit vom muskulären Ermüdungsgrad auf.

Demzufolge führt eine Dekonditionierung der Rückenmuskeln ebenfalls zu Koordinationsstörungen und muss somit zumindest als Risikofaktor für die Pathogenese des CURS berücksichtigt werden.

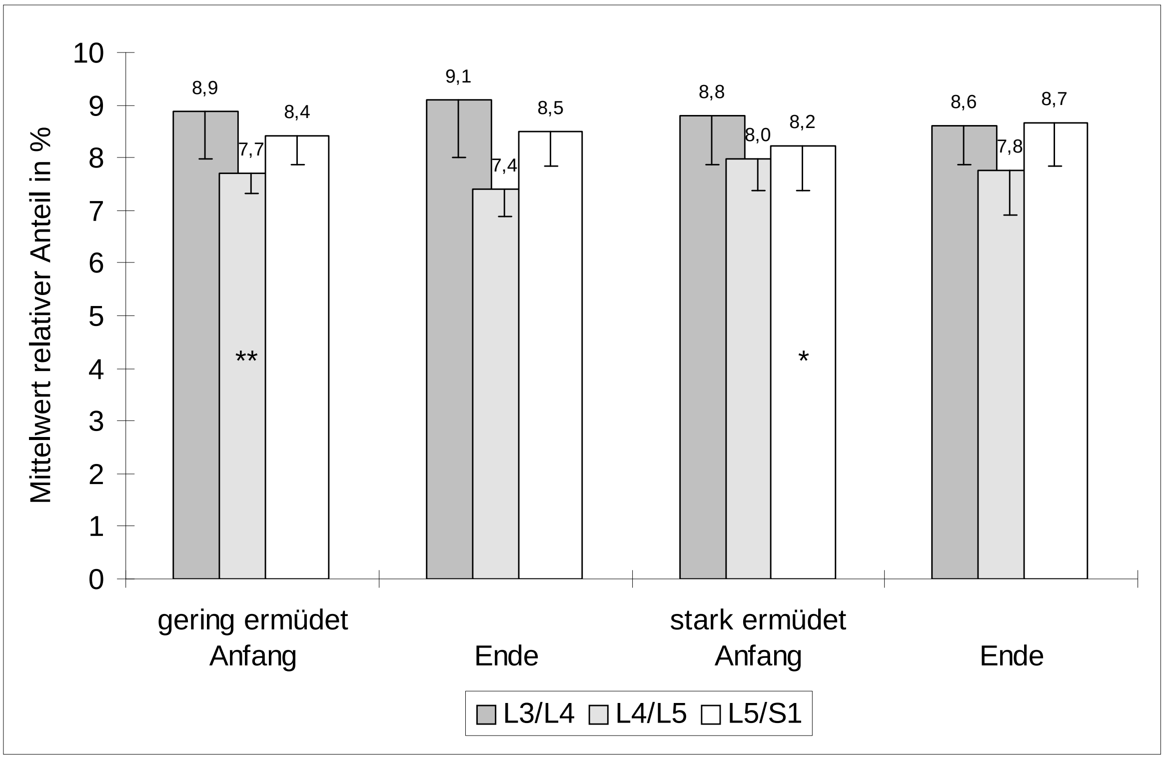


**Abbildung 1.1 Über den gesamten Extensionszyklus gemittelte relative Amplituden. Die Sterne in den Säulen markieren signifikante Unterschiede für den Vergleich Anfang gegen Ende der gesamten Untersuchung (** p<0,025, * p<0,05). Am Ende der Belastungssituation unterscheiden sich die Aktivitätsniveaus signifikant für die Gruppe der gering ermüdeten Personen (Zweiseitige Varianzanalyse, p<0,05). Die Fehlerindikatoren markieren die Standardabweichung der angegebenen Mittelwerte. Abbildung entnommen aus Anders, 1998 (Anders et al. 1998).**

## 1.3 Verwendetes diagnostisches Verfahren

Der diagnostische Zugang zur Problematik des CURS lässt sich nach übereinstimmender Meinung in der Literatur nicht durch radiologische Verfahren herstellen (Phillips et al. 1986, van Tulder et al. 1997): die angeschuldigten Mikrotraumen entziehen sich praktisch vollständig der verfügbaren bildgebenden Diagnostik. Hinweise ergeben sich jedoch aus der möglichen Diagnostik hinsichtlich auftretender Seitendifferenzen im Faserquerschnitt der Rückenmuskeln (Hides et al. 1994), bzw. dem Nachweis von Fetteinlagerungen in den entsprechenden Muskeln (Mooney et al. 1997). Dabei wies die Muskulatur der durch die Schmerzen betroffenen Seite einen geringeren Muskelquerschnitt auf (Hides et al. 1994). Unabhängig von der Seitenlokalisation fand sich bei den CURS-Patienten eine stärkere Fetteinlagerung als bei Gesunden (Mooney et al. 1997). Neuerdings ergeben sich auch Hinweise aus der Feindiagnostik der Facettgelenke, deren Fehlstellung, als so genannter artikulärer Tropismus

bezeichnet (Bogduk et al. 2000), zur Entstehung von Rückenschmerzen beitragen könnte (Jinkins 2004).
Neben der ebenfalls weit verbreiteten Anwendung bewegungsanalytischer Verfahren (Porter und Wilkinson 1997, Esola et al. 1996), eignet sich die Elektromyographie (EMG) sehr gut, um insbesondere den sensiblen Bereich gestörter muskulärer Funktion im Zusammenhang mit dem Auftreten von CURS zu evaluieren.
Dabei werden beide üblichen Verfahren, sowohl die Nadel-Elektromyographie (NEMG) als auch die Oberflächen-Elektromyographie (OEMG) erfolgreich eingesetzt.
Die initialen Befunde im Zusammenhang mit CURS wurden mittels invasiver EMG-Technik unter Verwendung von flexiblen Drahtelektroden gemessen (Hodges und Richardson 1996). Im klinischen Einsatz ist es jedoch schwierig den CURS-Patienten derartige invasive Methoden zu vermitteln. Des weiteren wird aufgrund der durch die geringe Ableitfläche bedingten Selektivität die Reliabilität von NEMG-Untersuchungen als eher gering eingeschätzt (Marshall und Murphy 2003).
Unter Berücksichtigung der möglichen Aussagen, die mittels OEMG getroffen werden können eignet sich jedoch die OEMG sehr gut für den Einsatz im Rahmen der Diagnostik des CURS. Die Hauptaussagen, die bei der herkömmlichen Anwendung der OEMG im bipolaren Verschaltungsmodus getroffen werden können beinhalten folgende Punkte:

- Analyse der muskulären Koordination
- Analyse des muskulären Ermüdungsverhaltens
- Analyse des muskulären Beanspruchungsgrades

Die Analyse der muskulären Koordination beinhaltet vor allem Aussagen hinsichtlich des zeitlichen Aktivierungsverhaltens der jeweils untersuchten Muskeln, wie sie beispielsweise bereits seit längerem bei der Ganganalyse zum Einsatz kommt (Joseph 1968, Dubo et al. 1976). Zusätzlich dazu beinhaltet die Koordinationsanalyse von Muskeln aber auch die Evaluierung der Aktivitätsverhältnisse verschiedener untersuchter Muskeln zueinander (van Dieen et al. 2003b).
Muskuläre Ermüdung ist definiert als temporärer Kraftverlust von Muskeln als Folge von muskulärer Tätigkeit (Gandevia et al. 1995b). Hierbei werden zwei

Formen unterschieden: zentrale Ermüdung und periphere Ermüdung (Gandevia et al. 1995a). Bei der zentralen Ermüdung lässt die Aktivierungsintensität seitens der zentralnervös initiierten Ansteuerung der Motoneurone nach (Loscher und Nordlund 2002), während bei der peripheren Ermüdung die Energiereserven der ausführenden Muskulatur erschöpft sind (Bentley et al. 2000). Beide Formen überlappen sich in der Praxis häufig (Kent Braun 1999). Insbesondere der Anteil der peripheren Ermüdung lässt sich anhand typischer OEMG-Veränderungen nachweisen (Luttmann et al. 1996, Dimitrova und Dimitrov 2003). Dabei steigt bei submaximalen Kontraktionsstärken die mittlere Signalamplitude an, während gleichzeitig die mittlere Frequenz absinkt (Luttmann et al. 1996). Ursache für diese Effekte sind hinsichtlich der Amplitudenveränderungen die über die Zeit zunehmende Rekrutierung zusätzlicher motorischer Einheiten (Westgaard und de Luca 1999, Jensen et al. 2000), sowie deren stärkere Synchronisierung (Grönlund et al. 2007, Kleine et al. 2001). Der Frequenzabfall wird ebenfalls durch die stärkere Synchronisierung (Kleine et al. 2001), aber auch durch die absinkende Ausbreitungsgeschwindigkeit der Muskelaktionspotentiale auf den Muskelfasern verursacht (Houtman et al. 2003).

Der aktuelle Beanspruchungsgrad von Muskeln wird anhand einer Referenzsituation determiniert. Hierzu dient als Goldstandard die Bestimmung der maximal möglichen willkürlich erzeugbaren Kraft, in den meisten Fällen als isometrische Maximalkraft (engl.: maximum voluntary contraction, MVC (Attebrant et al. 1995)). Die Bestimmung der isometrischen MVC unterliegt vielfältigen Einflüssen, wie Motivation (McNair et al. 1996) oder auch vorhandenen Schmerzen (Nicolaisen und Jorgensen 1985) um nur einzelne Beispiele zu nennen. Zudem ist es notwendig, für die Untersuchung mehrerer Muskeln, jeden Muskel einzeln zu testen, was neben dem erforderlichen Zeitaufwand die zu untersuchenden Personen erheblich beansprucht. Deswegen werden zunehmend Methoden der Normierung ohne die Applikation von MVC-Tests verwendet (Marras und Davis 2001, Mathiassen et al. 1995). Hierfür werden ebenfalls definierte, jedoch submaximale und somit weniger beanspruchende Tests verwendet.

## 1.4 Auswertungsparameter des OEMG

Die Aktivierung der motorischen Einheiten (ME) eines Muskels erfolgt nach dem Alles-oder-Nichts Prinzip (Basmajian und De Luca 1985), entsprechend der von den Motoneronen erzeugten Aktionspotenziale. Somit wird jede graduelle Ansteuerung von Muskeln als Resultat der Variation der Anzahl der aktiven ME, sowie ihrer Ansteuerungsfrequenz geregelt. Die Auswahl der anzusteuernden ME richtet sich grundsätzlich nach funktionellen Aspekten, deren wesentliche Komponenten die zu erzielende Kraft, sowie, insbesondere für flächige Muskeln, der Kraftvektor sind (Schumann et al. 1994). Dabei werden mit zunehmender Kraft zunächst kleine, vorwiegend aus Typ 1 Fasern bestehende ME eingesetzt (Kadefors et al. 1999). Bei zunehmender Kraftproduktion werden zunächst alle Typ 1 Fasern rekrutiert, bevor danach auch die Typ 2 Fasern angesteuert werden (Freund et al. 1975). Da die Ansteuerung der ME grundsätzlich funktionell determiniert ist, erfolgt ihre Aktivierung stochastisch (Basmajian und De Luca 1985). Bei einer EMG-Messung werden die Depolarisationen aller sich im Einzugsbereich der jeweiligen Messelektroden befindlichen Muskelfasern als Interferenzmuster erfasst (Basmajian und De Luca 1985). Dabei weist die Signalstärke eine Korrelation mit der Kontraktionsstärke auf, ist jedoch nicht mit dieser identisch (Lawrence und De Luca 1983). Die bisherige Forschung konnte die Art und Weise des Zusammenhanges in Abhängigkeit vom Änderungsmodus der beteiligten Mechanismen definieren: die Realisierung des Kraftanstieges durch Erhöhung der Feuerrate der beteiligten ME entsprach dabei einem linearen Amplituden-Kraft-Verhältnis, wohingegen bei einer zusätzlichen Rekrutierung weiterer ME eine S-förmige Abhängigkeit beider Parameter resultierte (Solomonow et al. 1990). Aufgrund der zugrunde liegenden Signalform als Dipol enthält das gemessene Interferenzmuster sowohl positive als auch negative Amplitudenwerte (Basmajian und De Luca 1985). Um die mittlere Amplitudenhöhe zu ermitteln macht sich somit eine Gleichrichtung des Interferenzsignals nötig. Der gebräuchlichste Parameter für die Bestimmung mittlerer Amplitudenwerte ist die so genannte "root mean square" (RMS), bei der die Betragsbildung durch Quadrierung der einzelnen Messpunkte erfolgt. Nach Mittelung der Einzelwerte über definierte Mittelungszeiträume erfolgt dann wieder die Berechung der Quadratwurzel (siehe Gleichung 1). Dieser Parameter wird für alle Analysen der vorliegenden Arbeit verwendet.

**Gleichung 1: Berechnungsalgorithmus der RMS, entnommen aus (De Luca und Knaflitz 1992)**

$$\mathrm{x}_{rms}(t) = \sqrt{\frac{1}{T}\int_0^T \mathrm{x}^2(t)dt}$$

Der Frequenzinhalt als zweite wesentliche Kenngröße des Interferenzsignals kann sehr genau durch die Fourieranalyse berechnet werden (Chatfield 1982, De Luca und Knaflitz 1992). Die Fourier-Analyse beschreibt das Zerlegen eines beliebigen periodischen Signals in eine Summe von Sinus- und Kosinusfunktionen (siehe Gleichung 2). Sie zerlegt ein komplexes Signal, wie es das OEMG darstellt damit in seine Frequenzanteile. Als Ergebnis sich anschließender Mittelungen erhält man als graphische Darstellung ein Periodogramm. Es enthält sowohl Informationen über die im Signal enthaltenen Frequenzbestandteile (*jx* in Gleichung 2) als auch über deren Amplitude ($A_j$ und $B_j$ in Gleichung 2). Somit können durch die Anwendung der Fourieranalyse sowohl Amplituden- als auch Frequenzparameter ermittelt werden (De Luca und Knaflitz 1992).

**Gleichung 2: genereller Berechnungsalgorithmus der Fourieranalyse, entnommen aus (Chatfield 1982, De Luca und Knaflitz 1992)**

$$f(x) = \frac{A_o}{2} + \sum_{j=1}^{\infty}\left(A_j * \cos(jx) + B_j * \sin(jx)\right)$$

Die Fourieranalyse wird vor allem für die Analyse muskulären Ermüdungsverhaltens verwendet, um gleichzeitig die Änderungen der Amplituden- und Frequenzwerte beurteilen zu können.

Beide genannten Methoden berechnen gemittelte Werte aus festgelegten Zeitabschnitten. Dabei gilt, dass entsprechend des stochastischen Charakters des zugrunde liegenden Interferenzsignals längere Abschnitte das Signal statistisch deutlich besser repräsentieren als kurze. Jedoch ist dabei zu beachten, dass für die betrachteten Analyseabschnitte das Signal Stationarität aufweisen muss (Chatfield 1982). Die mögliche Frequenzauflösung des Periodogramms entspricht dem Quotienten aus der aktuellen Abtastrate und der Anzahl der für die Berechnung verwendeten Werte (Chatfield 1982).

Als Abtastrate wird die Detektionsgeschwindigkeit der üblicherweise heutzutage verwendeten Computersysteme bezeichnet, mit der die mittels analoger Verstärkertechnik gemessenen Signale für die weitere Verarbeitung im Rechner digitalisiert werden. Daraus ergibt sich der minimal detektierbare Zeitabstand zweier aufeinander folgender Messpunkte. Für die vorliegenden Untersuchungen wurde immer mit einer Abtastrate von 2000/s gearbeitet, was in einem minimalen zeitlichen Detektionsabstand der Signale von 0,5 ms resultiert. Im Rahmen dieser so genannten analog-digital Wandlung muss für die mögliche Amplitudenauflösung der Signale die Anzahl der existierenden Messpunkte berücksichtigt werden. Diese Anzahl wird als Maßeinheit in bit (engl.: binary digit) angegeben. Dabei gilt, dass die dem bit vorangestellte Zahl die Anzahl der Messpunkte als Potenz von 2 entspricht. Die von uns in allen Studien verwendeten Messkarten hatten eine Amplitudenauflösung von 12 bit, was $2^{12}$ Werten, also 4096 Messpunkten entspricht. Für die in der Arbeit verwendeten Verstärkungsfaktoren resultiert das in einer minimal detektierbaren Amplitudenauflösung von 1µV.
Es werden noch eine Vielzahl weiterer Amplituden-, als auch Frequenzparameter berechnet, die, da sie in der weiteren Betrachtung der Arbeit keine Rolle spielen nicht weiter erwähnt werden. Detaillierte Angaben sind in der weiterführenden Literatur zu finden (De Luca und Knaflitz 1992).

# 2 Ziele der Arbeit

Die Arbeit stellt die Ergebnisse von Untersuchungen dar, deren Ziel die Analyse des Aktivierungsverhaltens von ausgewählten, repräsentativen Rumpfmuskeln unter definierten Provokationsbedingungen war. Dabei stand vor allem die Analyse der beobachteten Reaktionen im Hinblick auf die Einordnung der untersuchten Muskeln in die etablierte Systematik der Rumpfmuskeln im Vordergrund. Muskeln, die den jeweiligen funktionell definierten Muskelsystemen, also lokal, global stabilisierend oder global mobilisierend angehören (Comerford und Mottram 2001), sollten sich demnach innerhalb der Gruppen durch ähnliche Verhaltensweisen auszeichnen. Zwischen den Vertretern der verschiedenen Muskelsysteme sollten jedoch typische Unterschiede nachweisbar sein.

Die applizierten Belastungen erfassen für die untersuchten Muskeln beide Hauptaufgaben der Muskulatur: lokomotorische und haltemotorische Beanspruchungen.
Für die vorliegenden Untersuchungen wurden weiterhin graduell abgestufte Belastungsformen gewählt, um so insbesondere die Charakteristik der auftretenden Zusammenhänge zwischen der Änderung der Belastungsintensität in ihrer Auswirkung auf die Beanspruchung des muskulären Systems zu identifizieren.
Entsprechend der definierten Funktionscharakteristika (siehe Tabelle 1.1) können für die Muskelsysteme unterschiedliche Reaktionen auf die angebotenen Belastungen erwartet werden:

- lokale Muskeln: keine Änderung des Musters bei ansteigender Belastung/Belastungsart, lediglich moderate Amplitudenreaktion
- global stabilisierende Muskeln: Verstärkung vor allem exzentrischer Aktivierungen, stärkere Amplitudenreaktion bei zunehmender Belastungsintensität
- global mobilisierende Muskeln: Verstärkung vor allem konzentrischer Aktivierungen, starke Amplitudenreaktion bei zunehmender Belastungsintensität

Es wurden für alle Untersuchungen die gleichen Muskeln analysiert. Dabei handelte es sich um folgende Muskeln: M. rectus abdominis (RA, global mobilisierend), M. obliquus abdominis internus (OI), M. obliquus abdominis externus (OE, beide global stabilisierend), M. erector spinae, pars longissimus (ES, global mobilisierend) und den M. multifidus, pars lumbalis (MF, lokal stabilisierend). Alle Muskeln wurden immer simultan auf beiden Körperhälften gemessen.
Es wurden sowohl dynamische, als auch statische Untersuchungen ausgewählt, die, obwohl unter Laborbedingungen erfasst insofern praxisnah sind, als handelsübliche Test- und Trainingsgeräte verwendet wurden. Bei Vorhandensein der jeweils verwendeten Testgeräte können sie somit nachvollzogen und jederzeit angewendet werden.
In der ersten Studie wurden Untersuchungen auf einem Laufband durchgeführt.

Dabei lag das Augenmerk insbesondere darauf, die während des Gehens identifizierbaren Aktivitätsmuster der untersuchten Rumpfmuskeln als Normwerte zu etablieren. Die applizierten Gehgeschwindigkeiten variierten zwischen 2 und 6 km/h, um neben der Analyse der damit verbundenen zyklischen Provokation der Rumpfmuskeln die Auswirkung sich graduell ändernder Anforderungen zu identifizieren. Die beobachteten Aktivitätsmuster der untersuchten Rumpfmuskeln wurden funktionell charakterisiert und insbesondere in Hinblick auf ihre Zugehörigkeit zu den verschiedenen Muskelsystemen innerhalb der etablierten Systematik der Rumpfmuskeln (Comerford und Mottram 2001) analysiert.

Als weitere dynamische Provokation der Rumpfmuskeln wurden Oszillationen zwischen 3,0 und 4,5 Hz über ein vor dem Körper mit beiden Händen gehaltenes Therapiegerät (Propriomed®) appliziert. Das Gerät besteht aus einem flexiblen Rundstahl, der in der Mitte des Gerätes über einen Griff, sowie an den Enden über verschiebbare Gewichte verfügt. Über die Veränderung der Position der Gewichte kann die Eigenfrequenz des Gerätes genau eingestellt werden. Für die Benutzung des Gerätes werden die Enden in Schwingungen versetzt, die alternierende, jeweils entgegen gesetzte Kraftwirkungen auf den Oberkörper bewirken. Das Ziel der Untersuchung war wiederum, die hervorgerufenen Aktivitätsmuster der untersuchten Rumpfmuskeln zu identifizieren. Die Auswirkung sich quantitativ ändernder Anforderungen anhand der applizierten Frequenzen, sowie qualitativ unterschiedlicher Beanspruchungen durch die Applikation verschiedener Schwingungsrichtungen des Gerätes auf das Aktivitätsverhalten der Rumpfmuskeln stand hierbei im Mittelpunkt.

In der dritten Studie wurden die genannten Rumpfmuskeln statischen Belastungen unterzogen.

Die Untersuchung wurde in einem Gerät durchgeführt (Centaur®), welches die jeweilige Belastung dadurch appliziert, dass die sich im Gerät befindliche Person gekippt wird. Die zu untersuchende Person ist im Gerät bis zur Hüfte fixiert, der Oberkörper wird frei gehalten. Die Person hat während der Untersuchung lediglich die Aufgabe, ihren Oberkörper in der Längsachse des Körpers zu stabilisieren. Die Untersuchung wurde mit dem Ziel durchgeführt, die OEMG-Amplituden – Kraft Beziehung der untersuchten Rumpfmuskeln zu identifizieren. Die Zugehörigkeit der Rumpfmuskeln zu den verschiedenen Muskelsystemen ließ dabei unterschiedliche Reaktionsweisen erwarten. Damit sollte insbesondere eine

Varianzaufklärung der bisher wenig erklärbaren Unterschiede im Amplituden – Kraft Verhalten verschiedener Muskeln des Menschen erfolgen (Lawrence und De Luca 1983).

# 3 Ergebnisse

## 3.1 Aktivitätsmuster von Rumpfmuskeln beim Gehen auf dem Laufband

### 3.1.1 Amplitudenverlauf über den normierten Schritt

Mit Ausnahme des RA wiesen alle untersuchten Rumpfmuskeln ein über die applizierten Geschwindigkeiten als stabil zu bezeichnendes Aktivierungsmuster auf (siehe Abbildung 3.1), d. h. die prinzipiellen Eigenschaften der Aktivierungsabfolge über den normierten Schritt änderten sich lediglich quantitativ. Mit steigender Geschwindigkeit war für alle Muskeln ein sich graduell erhöhendes Niveau der auftretenden Maximalamplitude zu verzeichnen. Das Verhalten der vorkommenden Minimalamplituden erscheint uneinheitlich. Die abdominalen Muskeln wiesen insbesondere während der Propulsionsphase des kontralateralen Beines Amplitudenspitzen auf. Für die beiden schrägen Bauchmuskeln war während der ipsilateralen Propulsionsphase ebenfalls ein Amplitudenpeak zu verzeichnen, der in seiner Ausprägung die Höhe der Spitzenamplitude während der kontralateralen Propulsion jedoch nicht erreichte. Diese Charakteristik konnte für den RA ebenfalls, jedoch erst für mittlere bis hohe Gehgeschwindigkeiten verzeichnet werden. Zu den jeweiligen Fersenaufsatzzeitpunkten wiesen alle ventralen Rumpfmuskeln Amplitudensenken auf. Im Gegensatz dazu konnte für beide untersuchten Rückenmuskeln immer zum Zeitpunkt des Fersenaufsatzes ein Amplitudenmaximum verzeichnet werden. Die Höhe beider Spitzenaktivierungen war für den MF praktisch gleich hoch. Diese Amplitudenspitzen zeigten jedoch für den ES eine deutliche Asymmetrie mit hohen Amplituden während des kontralateralen Fersenkontaktes und das niedrige Niveau während der Standphasen lediglich geringfügig übersteigende Amplitudenerhöhungen während des ipsilateralen Fersenkontaktes (Anders et al. 2007c).

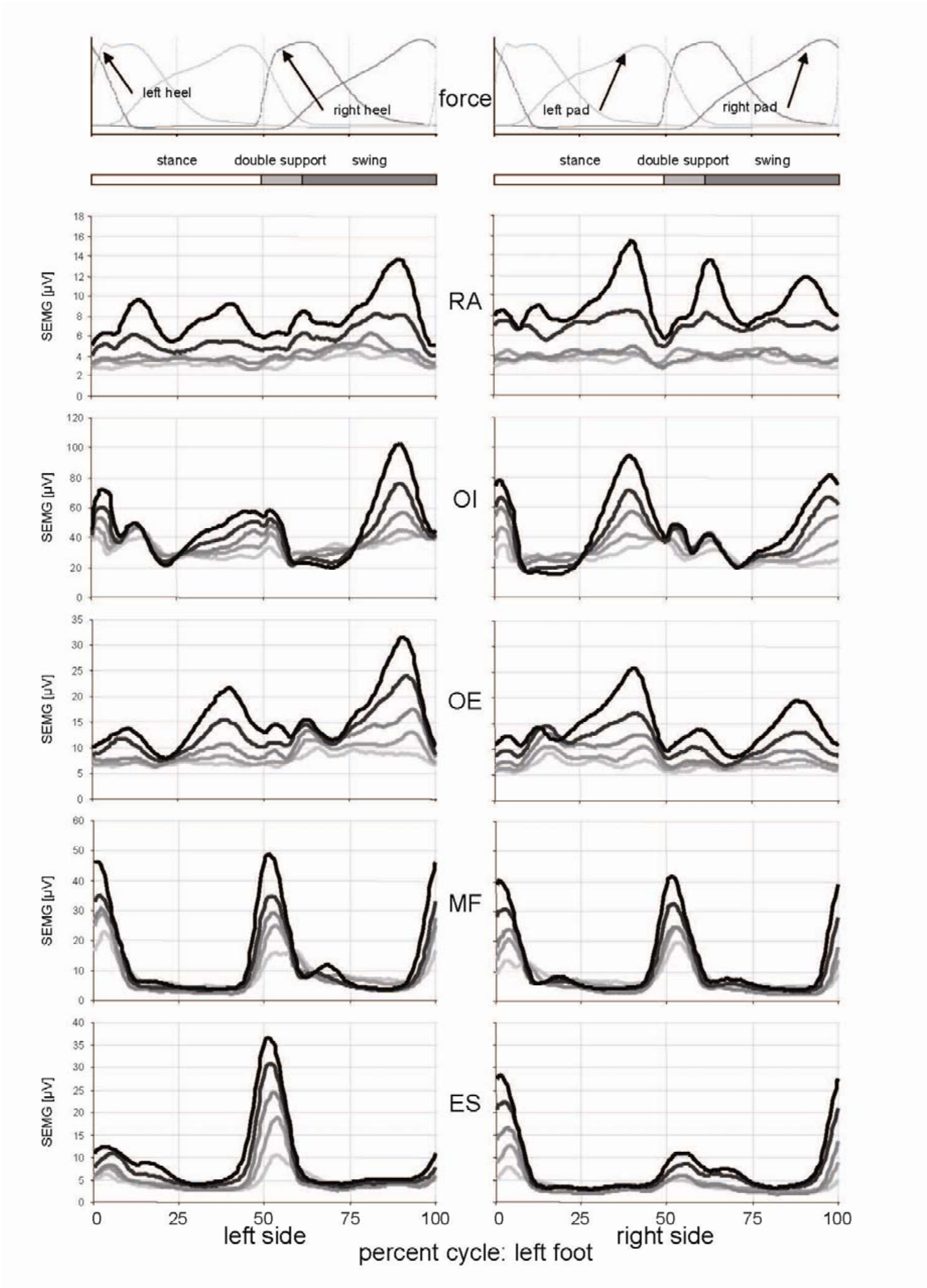


**Abbildung 3.1 Gemittelte OEMG-Kurven aller untersuchten Rumpfmuskeln. Die applizierten Geschwindigkeiten variieren zwischen 2 km/h und 6 km/h in Abstufungen von 1 km/h. Die dargestellten Grautöne werden mit steigender Geschwindigkeit dunkler. Der normierte Schrittzyklus wurde auf den Fersenaufsatz des linken Fußes normiert (siehe oberste Zeile). Abbildung entnommen aus (Anders et al. 2007c).**

### 3.1.2 Zeitunabhängige Amplitudenparameter über den normierten Schritt

Die mittleren Amplituden aller untersuchten Rumpfmuskeln stiegen mit steigender Gehgeschwindigkeit ebenfalls an. Das Bild für die im Schrittzyklus vorkommenden minimalen Amplitudenwerte war jedoch uneinheitlich: für den RA, OE und den ES stiegen diese Werte ebenfalls mit steigender Geschwindigkeit an, jedoch wies der OI für diesen Parameter einen Abfall mit sich erhöhender Gehgeschwindigkeit auf. Für den MF konnte eine "Senke" mit den niedrigsten Werten bei 4 km/h und jeweils ansteigenden Minimalamplitudenniveaus sowohl für niedrigere als auch höhere Gehgeschwindigkeiten beobachtet werden (siehe Abbildung 3.2). Die Varianz der Amplitudencharakteristik über den normierten Schritt kann durch die Anwendung des Variationskoeffizienten normiert und so direkt zwischen den verschiedenen Geschwindigkeiten verglichen werden. Dabei zeigte sich, dass hierbei das Verhalten der vorderen Rumpfmuskeln einheitlich ist: mit steigender Gehgeschwindigkeit stieg die Variabilität der Werte über den normierten Schritt ebenfalls an. Für die untersuchten Rückenmuskeln war das Verhalten ebenfalls einheitlich, jedoch davon abweichend: bis zu einer mittleren Gehgeschwindigkeit von 4 km/h stieg die Variabilität der Amplitudenwerte im Schritt ebenfalls an, um danach auf diesem Niveau zu verharren.

## 3.2 Aktivitätsmuster von Rumpfmuskeln während zyklischer Provokation mittels Propriomed®

### 3.2.3 Analyse des Amplitudenverhaltens der untersuchten Rumpfmuskeln

Die Oszillationsfrequenz konnte nur für den RA und den OE sich ändernde mittlere Amplitudenniveaus hervorrufen: die Amplitudenniveaus stiegen mit zunehmender Oszillationsfrequenz an. Eine Auswirkung auf das mittlere Amplitudenniveau durch geänderte Schwingungsrichtung des Gerätes war nur in einem Muskel, dem MF nachweisbar. Er wies für die horizontale Schwingungsrichtung eine niedrigere mittlere Amplitude auf als für die vertikale Schwingungsrichtung.

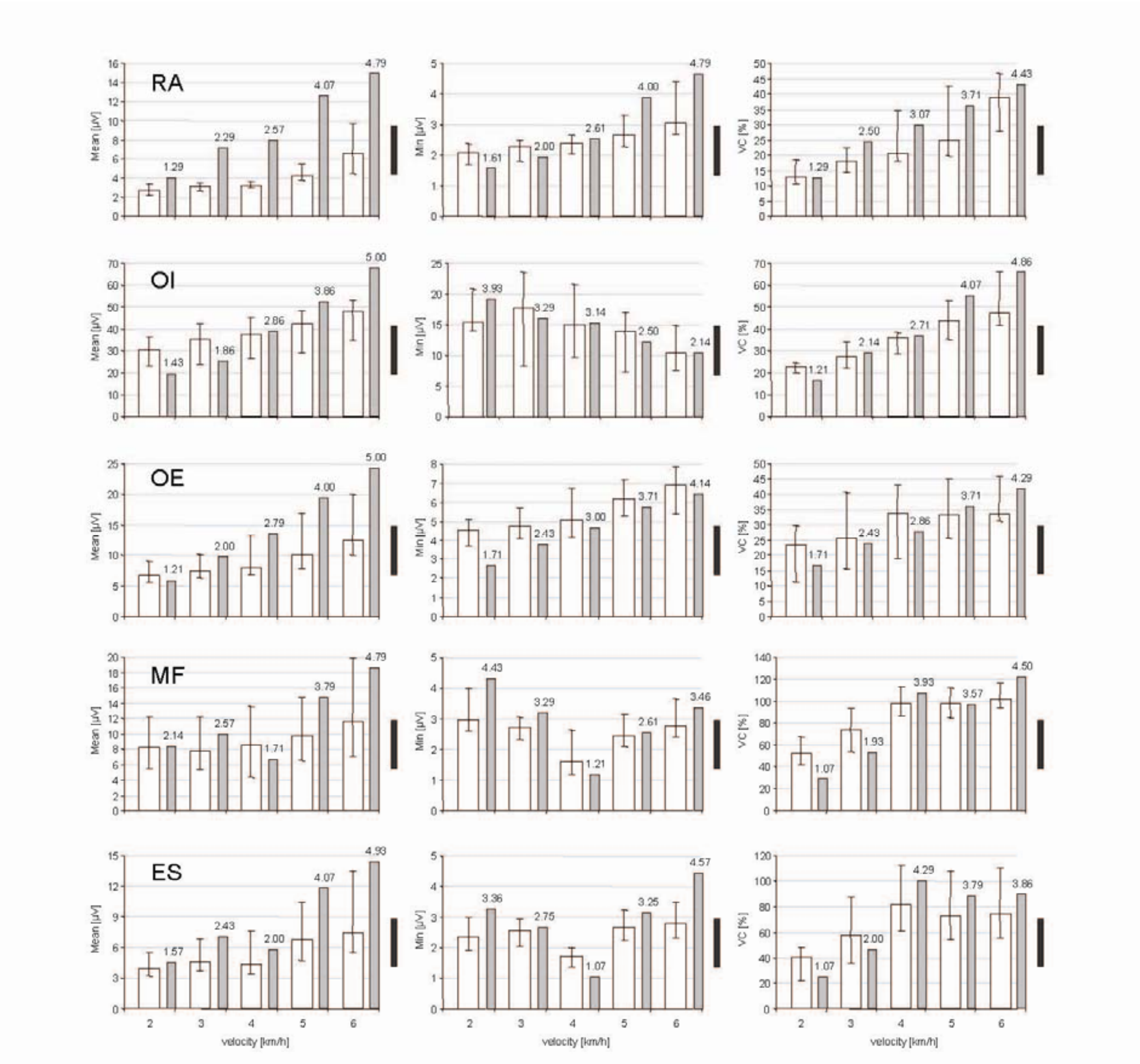


**Abbildung 3.2 Zeitunabhängige Parameter für alle applizierten Gehgeschwindigkeiten. Linke Spalte: mittlere Amplitude, mittlere Spalte: minimale Amplitude, rechte Spalte: Streuung der Amplitudenwerte als Variationskoeffizient. Die ungefüllten Säulen geben die Mediane, die Streuungsbalken die Abstände für das 1. und 3. Quartil wieder. Die schmalen gefüllten Säulen repräsentieren die mittleren Rangzahlen als Ergebnis des Friedman-Tests. Die Werte der mittleren Rangzahlen selber sind über den Säulen als Zahlenwert angegeben. Die rechts neben den Diagrammen positionierten Balken repräsentieren das kritische Signifikanzniveau von $p<0{,}05$, der kritische Rangzahlabstand beträgt 1,60 (Friedman-Test für abhängige Stichproben). Entnommen aus (Anders et al. 2007c).**

Der Amplitudenverlauf im zeitlich normierten Schwingungszyklus der untersuchten Rumpfmuskeln wurde durch die applizierten Variationen in der Oszillationsfrequenz und der unterschiedlichen Schwingungsrichtungen lediglich für die Rückenmuskeln beeinflusst. Alle ventralen Rumpfmuskeln behielten unabhängig von den genannten Einflussfaktoren ihre tonische Aktivierungscharakteristik bei. Beim Vergleich der Schwingungsrichtung wurde durch die vertikale Oszillation eine phasische Aktivierung beider Rückenmuskeln provoziert, während sie, wie ebenfalls für die restlichen Rumpfmuskeln bereits genannt, ein tonisches Amplitudenverhalten während horizontaler Schwingungsrichtung aufwiesen. Weiterhin verlagerte sich in Abhängigkeit von der applizierten Oszillationsfrequenz der auftretende Amplitudengipfel innerhalb des Schwingungszyklus: mit steigender Schwingungsfrequenz trat der Amplitudengipfel immer zeitiger auf. (Anders et al. 2008a), siehe Abbildung 3.3).

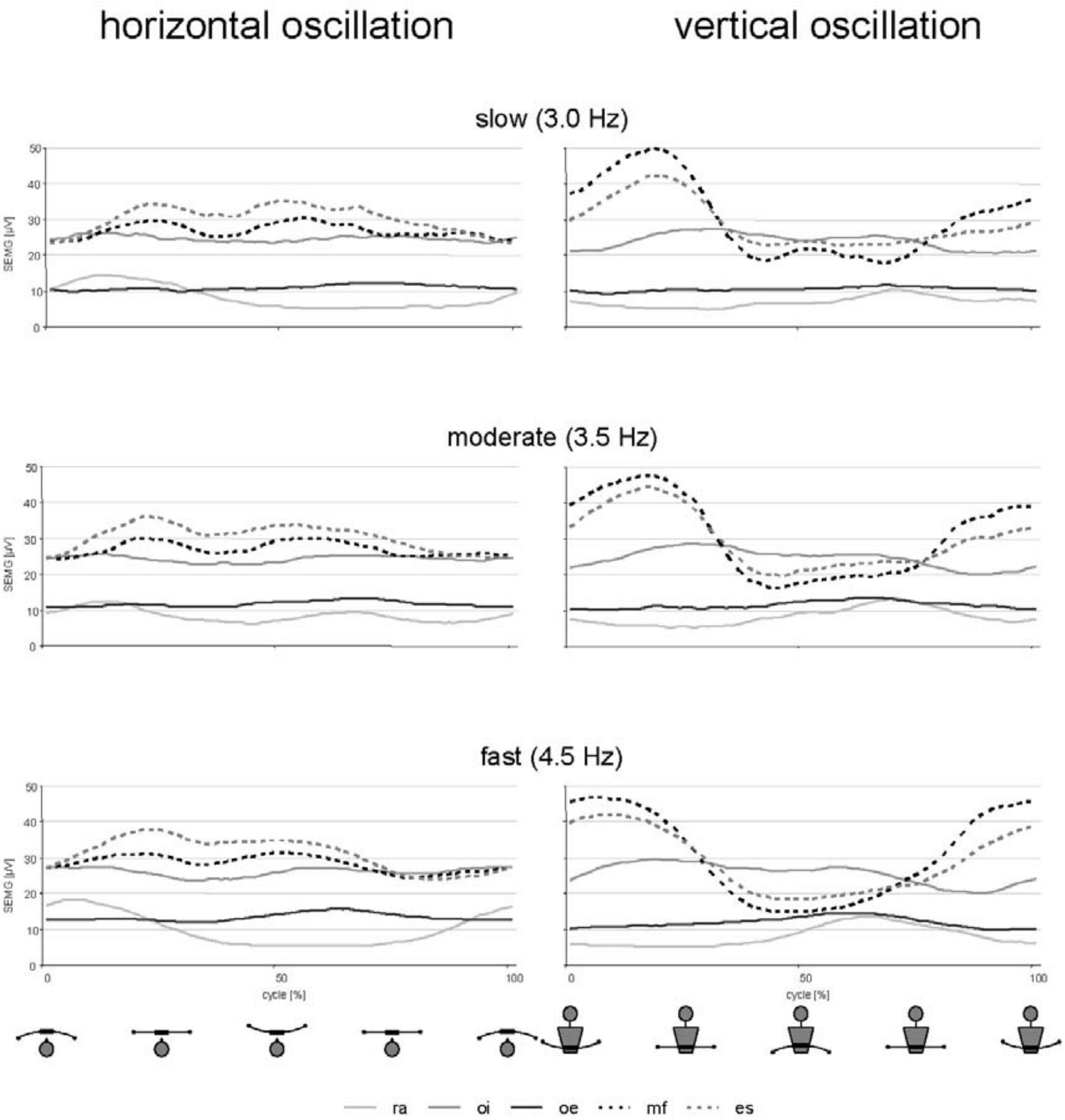


**Abbildung 3.3 Gemittelte OEMG-Kurven der untersuchten Rumpfmuskeln über den normierten Schwingungszyklus für die applizierten Frequenzen. Linke Spalte: horizontale Schwingungen. Rechte Spalte: vertikale Schwingungen. In der untersten Zeile ist die Position der Geräteenden während des normierten Schwingungszyklus schematisch angegeben. Abbildung entnommen aus (Anders et al. 2008a).**

### 3.2.4 Analyse des Koordinationsverhaltens der untersuchten Rumpfmuskeln

Für die Analyse der intermuskulären Koordination wurden wieder sowohl zeitunabhängige Parameter als auch zeitabhängige Parameter bestimmt. Für die Bestimmung der zeitunabhängigen Parameter wurden die zuvor gemessenen mittleren OEMG Amplituden als Relativamplituden, also prozentuale Anteile jedes einzelnen Muskels und der kumulativen Gesamtamplitude berechnet. Des weiteren wurden in Anlehnung an die Literatur so genannte Muskelratios (Edgerton et al. 1996, van Dieen et al. 2003b) als Verhältniswerte einzelner oder mehrerer Muskeln zueinander gebildet. Es wurden folgende Ratios berechnet: OI/OE, MF/ES und ventral/dorsal (in Abbildung 3.4 und Abbildung 3.5 als F/B bezeichnet). Diese Muskelratios wurden zudem nicht nur als zeitunabhängige Parameter, sondern auch für deren Verhalten als zeitabhängige Variable über den normierten Schwingungszyklus bestimmt.

Die zeitunabhängigen Parameter wiesen, wie bereits für die OEMG Amplituden nachgewiesen, eine Abhängigkeit des relativen Anteils des MF von der Schwingungsrichtung auf, der sich während vertikaler Schwingung gegenüber der horizontalen Oszillation erhöhte. Der Einfluss der Oszillationsfrequenz war zwar gering, jedoch vereinzelt ebenfalls nachweisbar. Während horizontaler Oszillationen unabhängig von der Schwingungsrichtung führte die Erhöhung der Schwingungsfrequenz zu einer Erhöhung der relativen Amplitude des OE und einer Abnahme des Anteils des MF. Eine Erhöhung des Anteils des RA konnte nur für die horizontalen Schwingungen nachgewiesen werden, während vertikaler Schwingungen verringerte sich der Anteil des ES mit steigender Schwingungsfrequenz, der des OI stieg demgegenüber an (Anders et al. 2007a).

Die aufgeführten Effekte ließen sich auch in den gebildeten Muskelratios nachweisen: sowohl die MF/ES als auch die ventral/dorsal Ratio wiesen während vertikaler Oszillation höhere Werte auf als während horizontaler Schwingungen. Die OI/OE Ratio wies zwar keinen Einfluss seitens der Schwingungsrichtung auf, während horizontaler Oszillationen verringerte sich jedoch ihr Wert mit zunehmender Schwingungsfrequenz signifikant. Ebenso unterlag die ventral/dorsale Ratio einem Einfluss der Schwingungsfrequenz, jedoch stieg hier der Wert mit ansteigender Schwingungsfrequenz während vertikaler Schwingung signifikant an (siehe Abbildung 3.4).

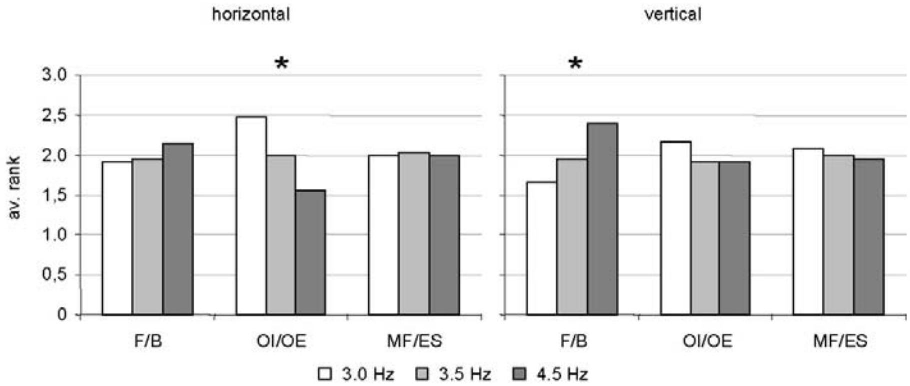


**Abbildung 3.4 Mittlere Rangzahlen für die Werte der mittleren berechneten Muskelratios in Abhängigkeit von der Schwingungsfrequenz. Durch die Sternchen sind signifikante Einflüsse der Schwingungsfrequenz gekennzeichnet ($p<0,05$, Friedman-Test). Abbildung entnommen aus (Anders et al. 2007a).**

Das Verhalten der berechneten Muskelratios als zeitabhängige Parameter erbrachte den generellen Effekt stärker alternierender Werte während vertikaler Schwingungsrichtung. Dieser Effekt war für die überwiegende Zahl der berechneten Werte signifikant (siehe Abbildung 3.5). Des weiteren konnte für die ventral/dorsal Ratio eine Umkehr im berechneten Amplitudenverhältnis nachgewiesen werden: während horizontaler Schwingung trat der Maximalwert unmittelbar vor dem hinteren Umkehrpunkt der Geräteenden auf, während für die vertikalen Schwingungen der Maximalwert direkt vor dem unteren Umkehrpunkt der Geräteenden zu verzeichnen war (siehe Abbildung 3.5).

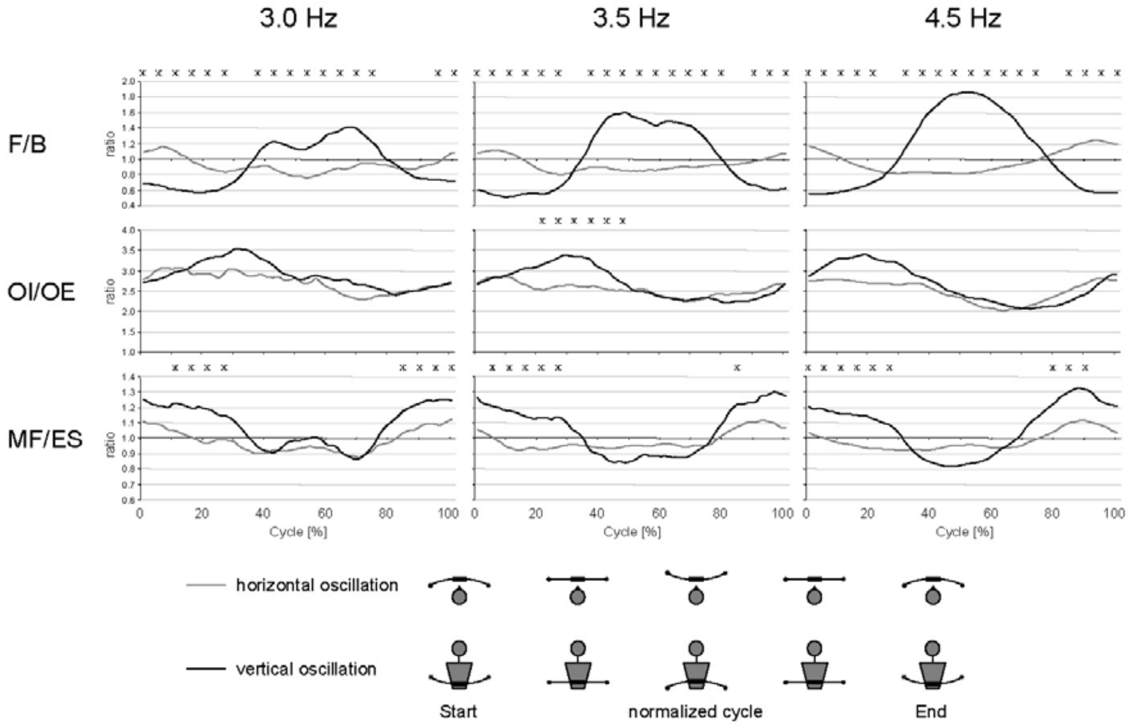


**Abbildung 3.5 Gemittelte Zeitverlaufskurven der berechneten Muskelratios über den normierten Schwingungszyklus. Sternchen markieren signifikante Unterschiede zwischen den beiden Schwingungsrichtungen (p<0,05, Wilcoxon-Test). Die Position der Geräteenden während des normierten Schwingungszyklus ist angegeben. Abbildung entnommen aus (Anders et al. 2007a).**

## 3.3 Identifikation der Amplituden – Kraft Beziehung von Rumpfmuskeln

Alle bisher genannten Rumpfmuskeln wurden in einer weiteren Studie statischen Belastungen unterzogen, um deren OEMG Amplituden – Kraft Beziehung zu untersuchen. Als wesentliches Ergebnis dieser Studie ist zu nennen, dass die untersuchten Rumpfmuskeln keine Amplituden – Kraft Verhältnisse aufwiesen, die ihrer Zuordnung zu den funktionell definierten Muskelsystemen (Comerford und Mottram 2001) entsprachen. Im Gegensatz dazu ergab sich eine lokalisationsbezogene Abhängigkeit: die ventralen Rumpfmuskeln wiesen eine nicht lineare Kurvencharakteristik auf, im Gegensatz dazu reagierten die untersuchten Rückenmuskeln mit einer praktisch ideal linearen Amplituden – Kraft Beziehung (siehe Abbildung 3.6) für die Belastungen in der Sagittalebene.

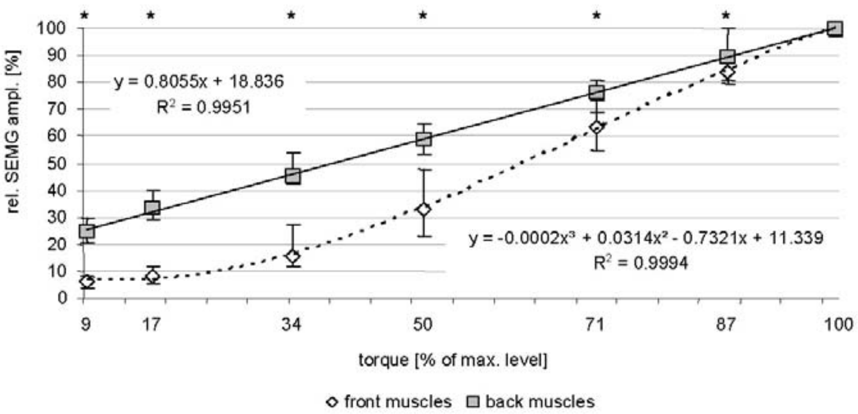


**Abbildung 3.6 Amplituden – Kraft Beziehung der gepoolten ventralen (während Rückkippung) und dorsalen (während Vorkippung) Rumpfmuskeln. Die Werte wurden auf das vorkommende Maximum normiert. Die dargestellten Werte entsprechen dem Median sowie den zugehörigen Quartilen. Die Sternchen markieren signifikante Differenzen zwischen den Werten ($p<0{,}05$, Wilcoxon-Test). Abbildung entnommen aus (Anders et al. 2008b).**

Die applizierten Belastungen in der Frontalebene bewirkten für alle untersuchten Muskeln geringere Amplitudenniveaus als für die Vor- und Rückkippung (siehe Abbildung 3.7). Für alle ventralen Rumpfmuskeln verringerte sich der Grad der Nichtlinearität. Die Rückenmuskeln reagierten auf den geänderten Kraftvektor mit leichten Veränderungen der Kurvencharakteristik hin zu ebenfalls nicht linearem Verhalten.

Die Unterteilung des untersuchten Probandengutes hinsichtlich ihrer Geschlechtszugehörigkeit erbrachte für die untersuchten Rückenmuskeln keine Unterschiede in der Kurvenform. Die Bauchmuskeln wiesen jedoch für die Gruppe der untersuchten Frauen gegenüber den untersuchten Männern eine weniger stark ausgeprägte Nichtlinearität auf (siehe Abbildung 3.8). Dieser Befund konnte allerdings lediglich für die Kippungen in der Sagittalebene nachgewiesen werden.

Die gemessenen OEMG-Amplituden wiesen ebenfalls systematische Geschlechtsunterschiede auf: während der Rückkippung zeigten die weiblichen Personen höhere OEMG Amplituden für alle Bauchmuskeln. Die Männer waren durch gegenüber den Frauen erhöhte Amplitudenwerte für den MF während der Vorkippung gekennzeichnet. Der ES wies eine Tendenz zu höheren Werten bei den Frauen auf (die tabellarische Auflistung aller Einzelwerte ist im Abschnitt 11 zu finden (Anders et al. 2007b))

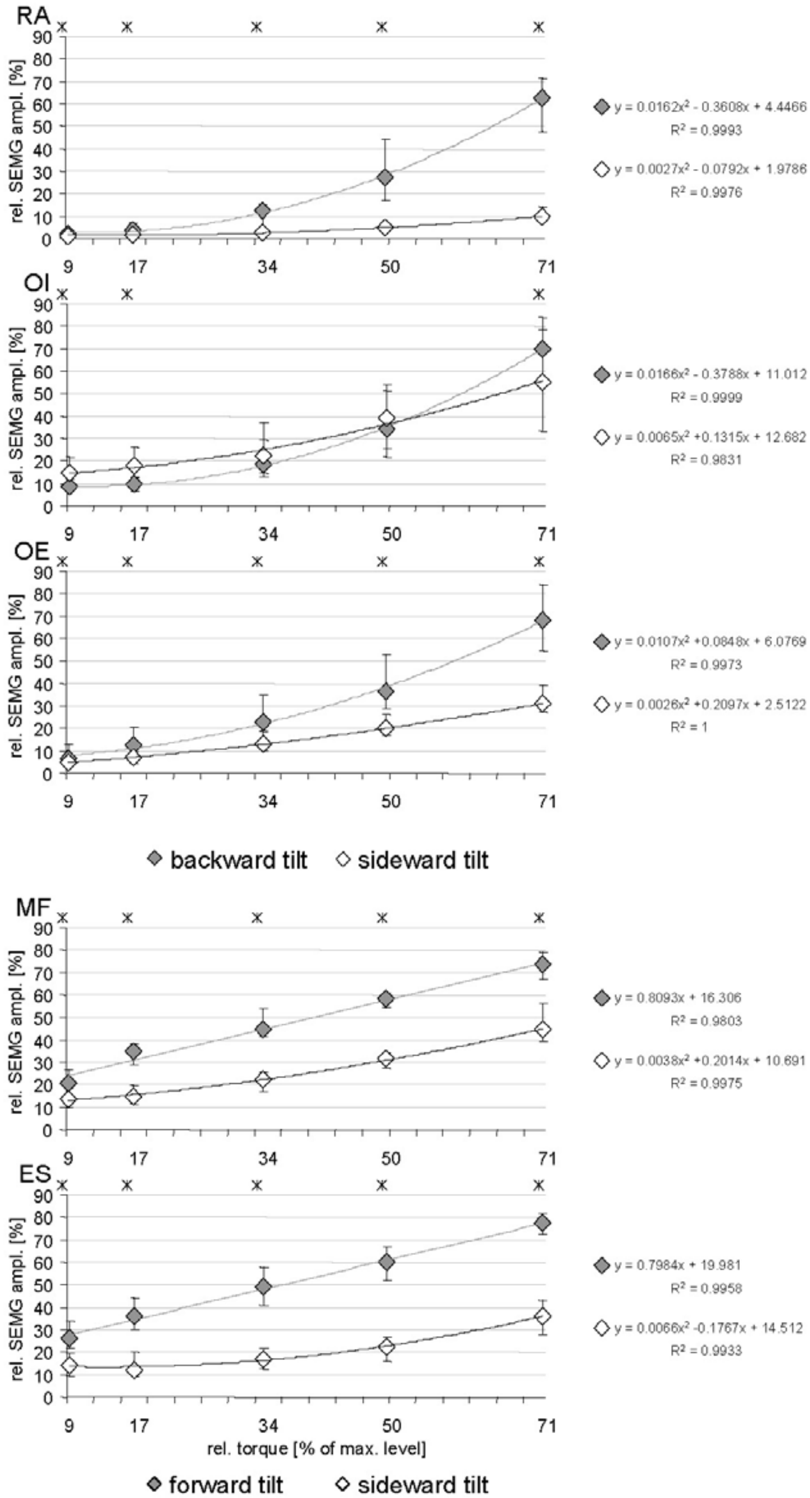


**Abbildung 3.7 Vergleich der auf den Wert bei 90° Kippung normierten Amplitudenwerte zwischen korrespondierenden Kippwinkeln in der Frontalebene und der Sagittalebene. Die Werte sind als Mediane und Quartile dargestellt. Die Sternchen repräsentieren signifikante Unterschiede zwischen den Kippebenen ($p<0,05$, Wilcoxon-Test). Abbildung entnommen aus (Anders et al. 2008b).**

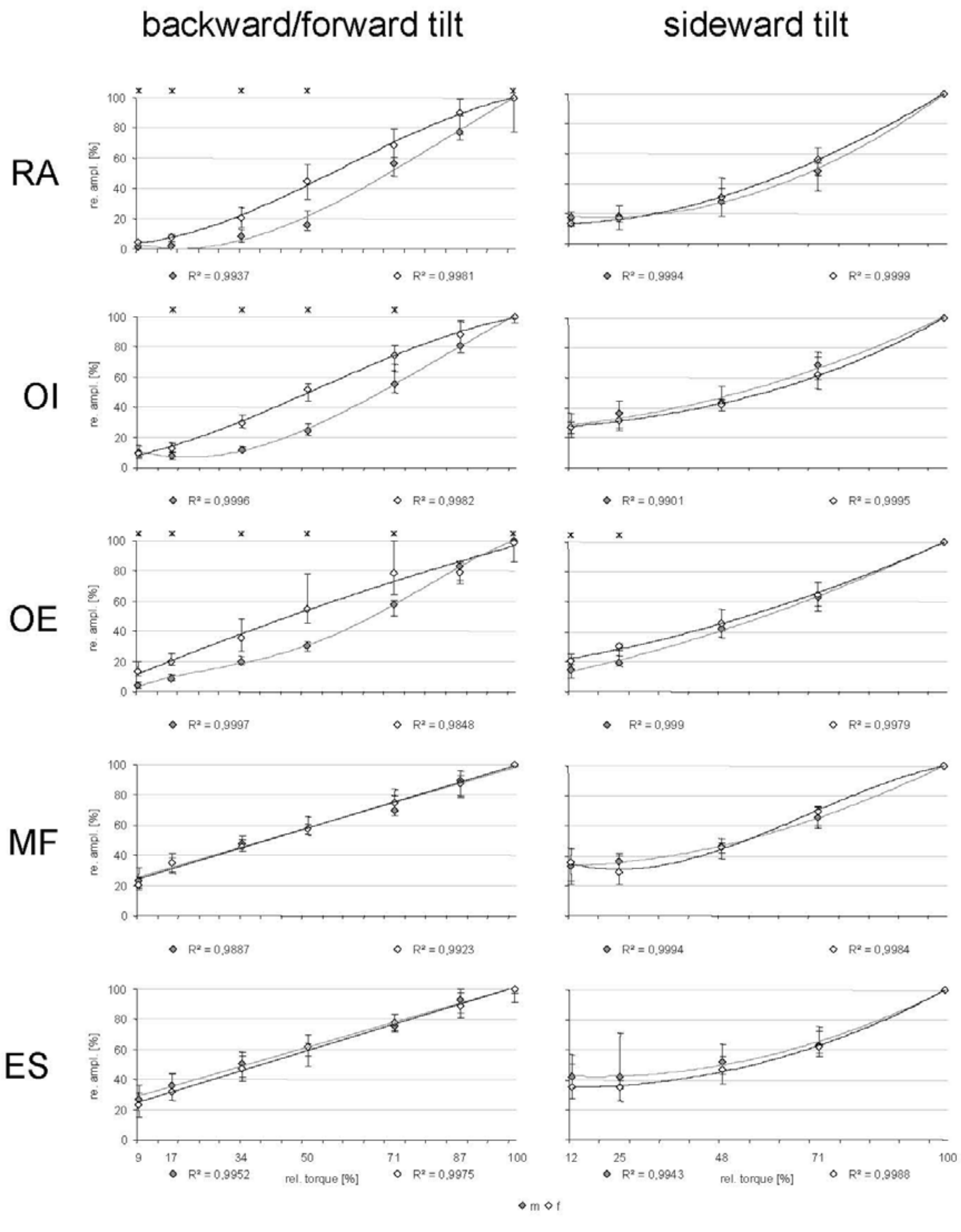


**Abbildung 3.8 Vergleich der normierten OEMG-Kurven zwischen Frauen und Männern. Die Werte repräsentieren Mediane und Quartile. Die Sternchen markieren signifikante Niveauunterschiede zwischen den Geschlechtern (p<0,05, U-Test). Abbildung entnommen aus (Anders et al. 2007b).**

## 4 Zusammenfassung

Die dargestellten Ergebnisse beinhalten verschiedene Ebenen provozierter muskulärer Aktivität.

Während des Gehens agiert die Rumpfmuskulatur unwillkürlich im Dienste der Lokomotion. Dabei handelt es sich um zyklische Aktivitätsmuster unterschiedlicher Frequenzen, bedingt durch die Spannbreite der applizierten Gehgeschwindigkeiten. Diese Muster stehen im Dienste der lokomotionsbedingten Rumpfbewegungen, die vor allem durch die zum Schritt gegenläufige Rumpfrotation und dem zugehörigen Armschwung gekennzeichnet sind. Gleichzeitig besteht jedoch die Aufgabe der Rumpfmuskulatur darin, während des Gehens die aufrechte Körperhaltung zu sichern.

Um die Daten im Weiteren in ihrer generellen Reaktionsweise besser interpretieren zu können, wurden die OEMG Verlaufskurven beider Seiten gepoolt. Beim ursprünglichen Bezug des normierten Schrittes auf den Fersenaufsatz des linken Beines wurden dafür die Daten des korrespondierenden rechten Muskels um 50% des normierten Schrittes verschoben. Damit bezieht sich das gepoolte Amplitudenverlaufsmuster nunmehr lediglich auf die Beziehung zum ipsilateralen bzw. kontralateralen Fersenaufsatz. Die Vorgehensweise ist in Abbildung 4.1 schematisch dargestellt.

Innerhalb der einzelnen Schrittzyklen wechseln sich Phasen konzentrischer mit Phasen exzentrischer Muskelaktivitäten ab: während der Standphase werden aufgrund der gegenläufigen Rumpfrotation beide ipsilaterale schräge Bauchmuskeln gedehnt, in diesem Zeitraum auftretende Aktivitäten sind somit exzentrisch. Für die beiden untersuchten Rückenmuskeln und den geraden Bauchmuskel sind exzentrische Kontraktionen während der kontralateralen Standphase zu verzeichnen, da die kontralaterale Beckenhälfte während der Standphase absinkt. Die Zuordnung ist in Abbildung 4.2 schematisch dargstellt.

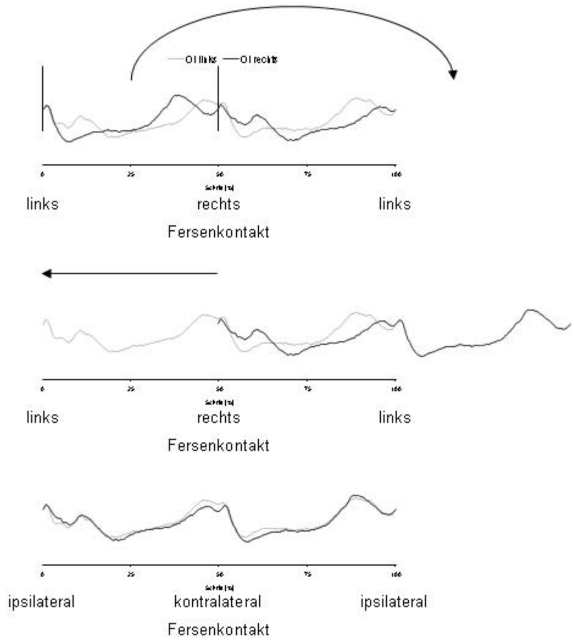


**Abbildung 4.1 Schematische Darstellung für die Vorgehensweise beim Poolen der OEMG-Verlaufskurven am Beispiel des OI für eine Gehgeschwindigkeit von 4 km/h. Bezugspunkt für die Schrittnormierung ist der linke Fersenaufsatz. Der rechte Muskel wird um 50% der Zykluslänge phasenverschoben. Die gepoolten Daten beziehen sich nur noch auf den ipsi- bzw. kontralateralen Fersenaufsatz**

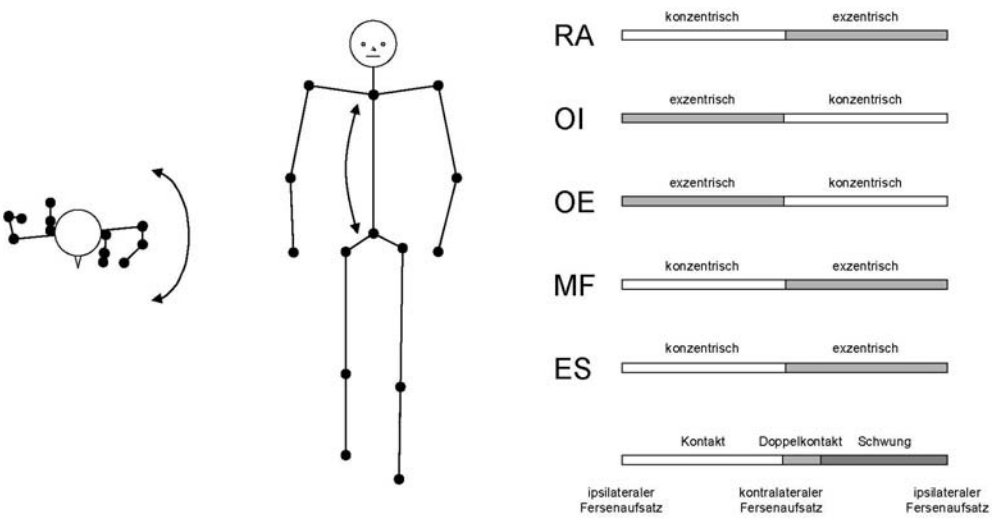


**Abbildung 4.2 Schematische Darstellung der zu erwartenden Kontraktionsformen für die untersuchten Rumpfmuskeln während Lokomotion. Die Pfeile symbolisieren die Regionen exzentrischer Muskelaktivität für die schrägen Bauchmuskeln (linkes Schema) und die geraden Rumpfmuskeln (rechtes Schema). In der rechten Spalte sind die zu erwartenden Kontraktionsformen in Bezug auf den normierten Schrittzyklus dargestellt.**

Obwohl praktisch alle untersuchten Muskeln keine grundsätzliche Änderung ihrer Aktivitätscharakteristik während der sich ändernden Gehgeschwindigkeiten zeigten konnte zumindest für die Rückenmuskeln die niedrigste mittlere Amplitude bei 4 km/h gemessen werden. Diese Geschwindigkeit entspricht ungefähr der zu veranschlagenden optimalen Gehgeschwindigkeit von gesunden Personen (Lamoth et al. 2005). Für alle Rumpfmuskeln nahm die Phasizität der Aktivierungsmuster über die beobachteten Geschwindigkeiten hinweg zu. Dabei wiesen die Rückenmuskeln wiederum den niedrigsten Wert des beobachteten Minimums bei 4 km/h auf, der OI war jedoch bis hin zur höchsten applizierten Gehgeschwindigkeit durch abfallende Minimalamplituden mit steigender Geschwindigkeit gekennzeichnet.

Die Analyse der Aktivitätsmaxima lässt weitere interessante Schlüsse zu (siehe Abbildung 4.3): der MF wies unabhängig von der Seitenlokalisation bei jedem Fersenkontakt ein Amplitudenmaximum auf. Damit zeigt er sowohl die erwartete exzentrische Kontraktion (Fersenaufsatz des kontralateralen Beines) als auch eine eher unerwartete konzentrische Kontraktion (Fersenaufsatz des ipsilateralen Beines). Dem ES fehlt während des ipsilateralen Fersenkontaktes diese Amplitudenspitze fast völlig. Damit würde dieser Muskel beim Gehen eher exzentrische Kontraktionen entwickeln.

Beide schrägen Bauchmuskeln waren grundsätzlich durch ein dreigipfliges Verhalten über den normierten Schritt gekennzeichnet: der prominenteste Gipfel während der kontralateralen Abdruckphase (konzentrisch). Während der ipsilateralen Abdruckphase ist der beobachtete Gipfel für den OI größer als für den OE (exzentrisch), sowie ein dritter Gipfel, kurz nach dem ipsilateralen Fersenaufsatz, der für den OE höher als für den OI ist (exzentrisch). Somit sind während der Propulsion immer die Muskeln beider Seiten aktiv: der ipsilaterale Muskel exzentrisch, der kontralaterale konzentrisch, jedoch ist der Anteil des OI während der exzentrischen Aktivitätsphase höher, bezogen auf die für jeden Muskel identifizierbare Maximalamplitude. Im Gegensatz zum OI kann für den OE des weiteren ein deutlich höherer, jedoch kurzer Aktivitätspeak kurz nach dem ipsilateralen Fersenaufsatz verzeichnet werden der ebenfalls exzentrisch ist.

Bei einer technisch realisierten Auflösung von 1µV für die verwendete Messkonfiguration ist die Rolle des RA trotz der deutlich ansteigenden Amplituden bei mittleren bis höheren Gehgeschwindigkeiten als vernachlässigbar einzuschätzen, da selbst die Spitzenwerte der gemessenen Amplituden gerade einmal 16 µV betragen. Der vor allem bei höheren Gehgeschwindigkeiten identifizierbare Amplitudengipfel ist jedoch als exzentrisch einzuordnen, was der postulierten Funktionscharakteristik widerspricht.

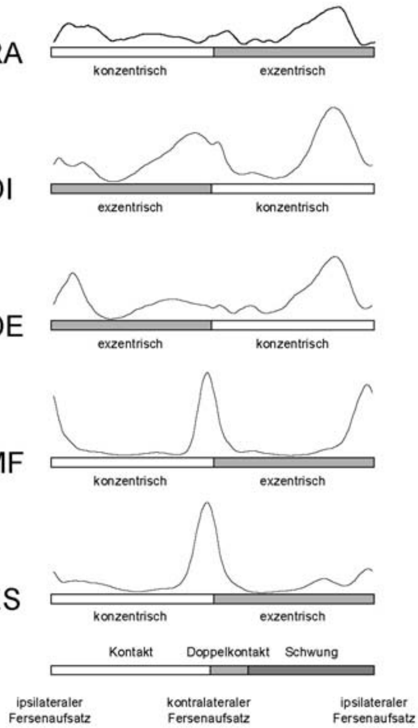


**Abbildung 4.3 Aktivitätsprofile der untersuchten Rumpfmuskeln in ihrer Beziehung zu den Schrittphasen und den dabei zu erwartenden Kontraktionsformen. Der besseren Übersichtlichkeit halber wurden die Daten beider Seiten gepoolt. Es sind die Muster während Gehens bei 6 km/h dargstellt.**

Es konnte zwar für die schrägen Bauchmuskeln als Vertreter des global stabilisierenden Systems sehr wohl eine stabilisierende Aktivitätscharakteristik identifiziert werden, jedoch wiesen beide ihre höchsten Amplituden während konzentrischer Kontraktionsphasen auf. Die untersuchten Rückenmuskeln zeigten ebenfalls unerwartete Aktivitätsmuster. Die Charakteristik des ES mit einem eingipfligen Muster, welches zudem an der Übergangsphase zur exzentrischen Belastung lokalisiert ist, wäre für diesen als einen typischen Vertreter der global mobilisierenden Muskeln nicht zu erwarten gewesen. Die doppelgipflige Antwort des MF erscheint hingegen wiederum genau der Charakteristik eines global stabilisierenden Muskels zu entsprechen. Da die Messung an der Oberfläche zwangsläufig Anteile des Muskels erfasst die mehrere Segmente überstreichen, muss aber davon ausgegangen werden, dass diese Anteile eher entsprechend global stabilisierender Aktivitätscharakteristika angesteuert werden. Angaben aus der Literatur unterstützen diese Sichtweise (Moseley et al. 2002). Insofern stellt der Befund für den MF keinen wirklichen Widerspruch zur postulierten Funktionsweise dar.

Der ständige Wechsel zwischen konzentrischen und exzentrischen Kontraktionen für beide schrägen Bauchmuskeln und den MF geben deutliche Hinweise darauf, dass nicht die Aktivitätscharakteristik allein sondern insbesondere das wechselnde Zusammenspiel der genannten Muskeln in fein koordinierter Abstimmung zueinander die Rumpfrotation ermöglicht und gleichzeitig die erforderliche Stabilität der WS sichert. Dabei spielt die Zuordnung der beobachteten Kontraktionen zu den Kontraktionsformen offensichtlich eine eher untergeordnete Rolle. Vielmehr bestehen während der Fersenaufsatzzeiten und den Phasen der Propulsion in ihrer Summation erhöhte Anforderungen an das Gesamtsystem: zum Fersenaufsatzzeitpunkt muss der plötzlich einwirkende Kraftvektor durch die Bremswirkung des aufsetzenden Fußes, sowie die geänderte Balancesituation kompensiert werden. Während der Propulsion muss die Rotationsbewegung des Rumpfes umgekehrt werden.

Befunde, die an Patienten mit CURS gewonnen wurden, konnten eine erhöhte Kokontraktionsaktivität für diese Patienten nachweisen (van Dieen et al. 2003b, McGill et al. 2003). Experimentell verursachte Schmerzen führten in einem weiteren Experiment für den ES und MF zu einer deutlichen Störung des im Normalfall fast invarianten Aktivierungsmusters aufeinander folgender Schritte (Lamoth et al. 2004). Im selben Experiment führte zudem bereits die Angst vor den zu erwartenden Schmerzen zu ähnlichen Abweichungen der Schritt – zu – Schritt Variationen wie die Schmerzen selber (Lamoth et al. 2004). Weiterhin konnten Auswirkungen auf die Bewegungsmuster des Rumpfes nachgewiesen werden (Lamoth et al. 2005).

Durch die Handhabung des Propriomed® wurden ebenfalls zyklische Kraftwirkungen auf den Rumpf appliziert. Diese Störungen wurden jedoch nicht durch eine gewohnte, im Unterbewusstsein generierte Zyklik (MacKay-Lyons 2002) verursacht, sondern durch die justierbare Eigenschwingungsfrequenz des Gerätes vorgegeben.

Versucht man erneut, die beobachteten Aktivitäten den provozierten Kontraktionsformen zuzuordnen, ergibt sich folgendes Bild (siehe Abbildung 4.4): im Bereich der Bewegungsumkehr für die rückwärts, bzw. aufwärts gerichteten Bewegungsphasen wäre eine exzentrische Kontraktion für die ventralen Muskeln zu erwarten. Den restlichen Bewegungsphasen wären demnach konzentrische Kontraktionen zuzuordnen. Die untersuchten Rückenmuskeln sollten sich in grober Näherung spiegelbildlich verhalten, jedoch kommt bei den horizontalen Schwingungen die zu verrichtende Haltearbeit aufgrund des Eigengewichtes des Gerätes hinzu.

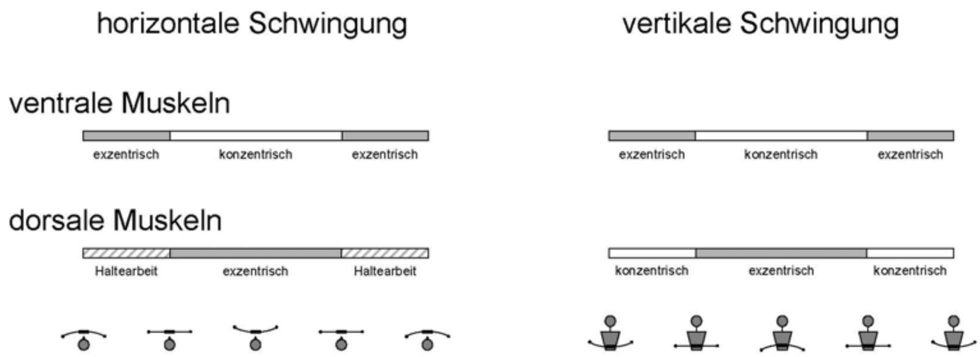


**Abbildung 4.4 Schematische Darstellung der erwarteten Aktivitätscharakteristik für die untersuchten Rumpfmuskeln während der Benutzung des Propriomed®. Linke Spalte: horizontale Schwingungsrichtung, rechte Spalte: vertikale Schwingungsrichtung.**

Die Rumpfmuskeln reagierten auf diese Provokation, wenn sie in horizontaler Schwingungsrichtung appliziert wurde mit einer fast unveränderlichen, tonischen Aktivierung. Die gering ausgeprägten Amplitudenmodulationen wiesen für die beiden schrägen Bauchmuskeln die höchsten Amplitudenwerte während exzentrischer Beanspruchung auf. Im Gegensatz dazu veränderten die Rückenmuskeln ihr Aktivierungsverhalten hin zu phasischer Aktivierung wenn vertikale Schwingungen appliziert wurden. Der auftretende Amplitudengipfel lag dabei kurz nach dem oberen Umkehrpunkt der Geräteenden. Unter Hinzurechnung der Zeit für die elektromechanische Ankopplung und der Trägheit des Oberkörpers (Thelen et al. 1994) erfolgt die mechanische Muskelaktion jedoch zum Zeitpunkt des unteren Umkehrpunktes (siehe Abschnitt 8). Somit weisen beide Rückenmuskeln wiederum exzentrische Kontraktionen auf (siehe Abbildung 4.5).

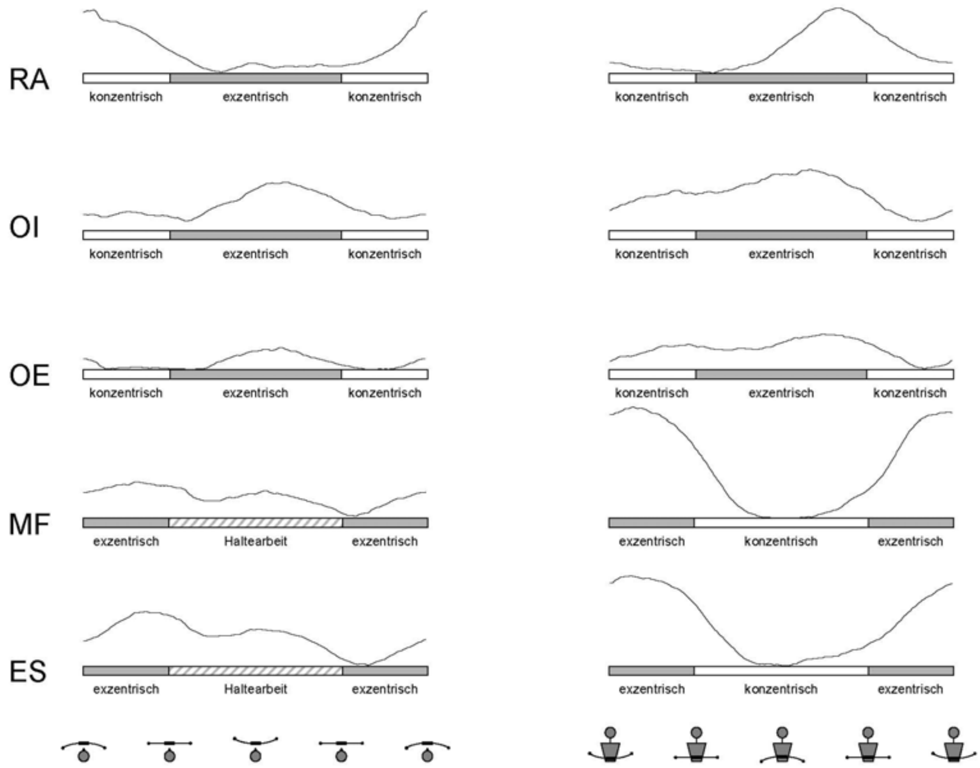


**Abbildung 4.5 Aktivitätsprofile der untersuchten Rumpfmuskeln in ihrer Beziehung zu den Bewegungsphasen des Propriomed® und den zu erwartenden Kontraktionsformen. Die Lage der erwarteten Kontraktionsformen wurde um ½ der Zykluslänge verschoben, um die zu berücksichtigende elektromechanische Ankopplungszeit zu kompensieren.**

Die applizierten Schwingungsfrequenzen hatten nur vereinzelt Einfluss auf die mittlere Amplitude aller Rumpfmuskeln, jedoch wurden die Amplitudenverhältnisse in zwei Fällen deutlich verändert. Während horizontaler Schwingungen verringerte sich mit steigender Schwingungsfrequenz die OI/OE Ratio systematisch. Dem gegenüber erhöhte sich die ventral/dorsale Ratio zusammen mit steigender Schwingungsfrequenz.

Somit blieben die Bauchmuskeln von der Art der applizierten Störung weitgehend unabhängig und dabei tonisch aktiv. Die Veränderung der Schwingungsrichtung von horizontal nach vertikal führte zu einer einheitlichen Reaktion beider untersuchten Rückenmuskeln hin zu phasischer Aktivierung.

Als wesentliches Ergebnis im Zusammenhang mit dieser Untersuchung ist also die unterschiedliche Reaktionsweise der ventralen und dorsalen Rumpfmuskeln zu nennen. Alle Bauchmuskeln, unabhängig davon ob sie dem global mobilisierenden

System (RA) oder dem global stabilisierenden System zuzuordnen sind (OI, OE) wiesen die gleiche tonische, unabhängig von der Schwingungsrichtung nachweisbare Aktivierung auf. Während der horizontalen Schwingung ist zwar von einer durch die Oszillation lediglich modulierten Anforderung der Rumpfmuskeln auszugehen, da ja das Eigengewicht des Gerätes dauerhaft wirkt, der Wechsel zur vertikalen Schwingung resultiert jedoch in einer phasenweise Verringerung bzw. Erhöhung des wirkenden Momentes. Dennoch änderten lediglich die Rückenmuskeln ihr Aktivierungsverhalten. Dies geschah einheitlich für beide untersuchten Muskeln und deshalb unerwartet.

Als dritte Provokation wurden statische Belastungssituationen gewählt. Auch in diesem Fall zeigte sich nicht wie zunächst erwartet ein Zusammenhang zwischen den beobachteten Amplituden – Kraft Verlaufskurven und der funktionellen Zuordnung der Muskeln zu den etablierten Muskelsystemen (Comerford und Mottram 2001), sondern erneut eine streng lokalisationsbezogene Abhängigkeit der identifizierten Kurvenform: die Rückenmuskeln wiesen eine praktisch ideal lineare Kurvenform auf, während die Bauchmuskeln ein nichtlineares, s-förmiges Verhältnis entwickelten. Diese Zusammenhänge waren jedoch auf die Kippungen in der Sagittalebene begrenzt. Ein Wechsel zu Kippungen in der Frontalebene führte neben deutlich geringeren Amplitudenwerten auch zu veränderten Kurvenformen, die für die untersuchten Bauchmuskeln weniger nichtlinear waren. Im Gegensatz dazu wichen nunmehr die Rückenmuskeln von ihrer Linearität ab und zeigten gering ausgeprägtes nichtlineares Verhalten. Die Unterteilung des Probandengutes in Männer und Frauen ergab als einzigen systematischen Unterschied eine deutlich weniger stark ausgeprägte Nichtlinearität für alle Bauchmuskeln während der applizierten Rückkippung.
Diese Befunde korrespondieren mit bekannten histomorphologischen Unterschieden zwischen Bauch- und Rückenmuskeln einerseits und geschlechtsspezifischen Unterschieden in der Faserquerschnittsfläche von Typ 1 und Typ 2 Fasern andererseits.
Obwohl von einer erheblichen interindividuellen Varianz ausgegangen werden muss, kann dennoch angenommen werden, dass der Typ 1 Faseranteil der Rückenmuskeln mit Werten zwischen 54%-65% (Johnson et al. 1973, Mannion 1999) deutlich höher ist als der der untersuchten Bauchmuskeln. Für die

Bauchmuskeln kann nur von einem wesentlich geringerem Anteil, von 46%-58% (Haggmark und Thorstensson 1979, Johnson et al. 1973) ausgegangen werden. Des Weiteren unterscheiden sich die Faserzusammensetzungen zwischen den Geschlechtern zwar nicht systematisch (Staron et al. 2000, Toft et al. 2003), Männer verfügen aber über eine größere Faserfläche von Typ 2 Fasern (Miller et al. 1993, Simoneau und Bouchard 1989). Die beobachteten Unterschiede in der Kurvenform zwischen abdominalen- und Rückenmuskeln einerseits und zwischen Frauen und Männern für die untersuchten Bauchmuskeln andererseits weisen somit eine verallgemeinerbare Tendenz auf: der Grad der Nichtlinearität der Amplituden – Kraft Kurve scheint mit abnehmendem Anteil an Typ 1 Querschnittsfläche für die untersuchten Muskeln zuzunehmen.

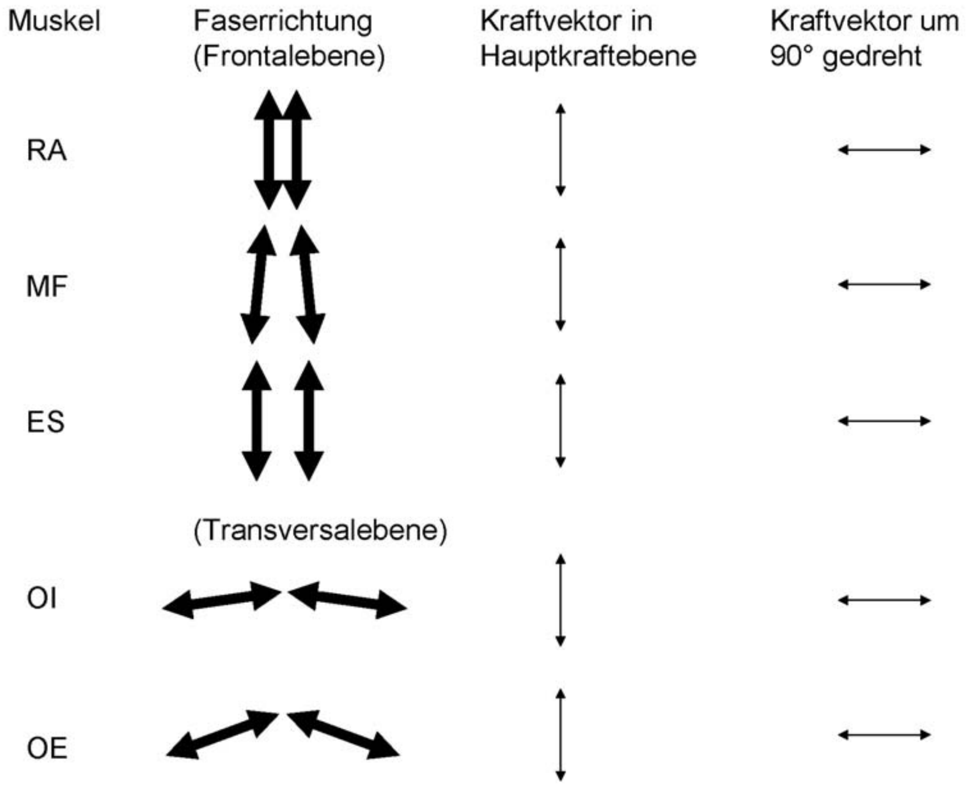


**Abbildung 4.6 Schematische Darstellung der hauptsächlichen Faserrichtung für die untersuchten Rumpfmuskeln in ihrer räumlichen Beziehung zum Kraftvektor während Ganzkörperkippung im Centaur®. Für die schrägen Bauchmuskeln wurde diejenige Faserrichtung angenommen, die für die verwendete Elektrodenlokalisation zu berücksichtigen ist.**

Ein Zusammenhang zwischen einwirkendem Kraftvektor und der identifizierbaren Kurvenform scheint ebenfalls zu existieren. Dabei könnte ein zum Faserverlauf parallel wirkender Kraftvektor eine grundsätzlich "linearisierende Wirkung" auf den Amplituden – Kraft Zusammenhang haben. Die Beziehungen zwischen den für die verwendeten Elektrodenpositionen anzunehmenden Faserrichtungen und den entsprechenden Kraftvektoren sind in Abbildung 4.6 dargestellt. Anhand der vorliegenden Daten erscheint also ein Zusammenhang zwischen Faserrichtung und einwirkendem Kraftvektor nahe liegend, kann aber, da diesbezüglich keine systematischen Untersuchungen vorgenommen wurden nicht verifiziert werden.

Experimentelle Befunde konnten zeigen, dass die Kurvenform des Amplituden – Kraft Zusammenhangs von der Ansteuerungscharakteristik des motorischen Systems abhängt (Solomonow et al. 1990): solange ein Kraftzuwachs lediglich über die Erhöhung der Feuerrate der ME bedingt ist konnte ein linearer Zusammenhang gefunden werden. Erst durch die zusätzliche Rekrutierung weiterer ME änderte sich die Kurvenform hin zu nichtlinearem Verhalten. Dies lässt sich üblicherweise vor allem während der Generierung höherer Kraftniveaus (>50% MCV) nachweisen. In der vorliegenden Untersuchung konnte ein solcher Zusammenhang jedoch nicht angenommen werden: die untersuchten Frauen waren durch die applizierte Belastung, vor allem für die Rückkippung deutlich stärker beansprucht als die untersuchten Männer. Demzufolge wäre für sie eine deutlichere Abweichung von der linearen Kurvenform zu erwarten gewesen. Dies trat nicht auf. Für die Form der OEMG - Antwortkurve auf statische Belastungen hat somit die Faserzusammensetzung in Wechselwirkung mit dem wirkenden Kraftvektor offensichtlich eine entscheidende Bedeutung.

Ein Zusammenhang zur unterschiedlichen Zuordnung der untersuchten Rumpfmuskeln zu den funktionell determinierten Muskelsystemen ließ sich erneut nicht feststellen.

Die derzeit international als verbindlich anerkannte Zuordnung der Rumpfmuskeln zu Muskelsystemen und damit die Bewertung ihrer funktionellen Ansteuerung muss vor dem Hintergrund der dargstellten Ergebnisse kritisch hinterfragt werden.

Während statischer Belastungen spielt für die Charakteristik des Amplituden - Kraft Zusammenhanges offensichtlich die Faserzusammensetzung und die Wechselwirkung zwischen dem jeweiligen Kraftvektor in Relation zur vorwiegenden Faserrichtung eines Muskels eine größere Rolle als bisher angenommen.

Die beobachtbaren Aktivitätsmuster während der unterschiedlichen zyklischen Provokationen sind streng an die jeweilige Belastung angepasst. Dabei können die untersuchten Muskeln sowohl tonisch als auch phasisch aktiviert werden, exzentrische Kontraktionen wechseln sich mit konzentrischen ab. Im untersuchten Bereich abgestufter Belastungsintensitäten wies letztlich jeder Muskel ein individuelles Reaktionsmuster auf, das wiederum, wie auch schon während der statischen Untersuchungen sichtbar geworden, unabhängig von seiner Zuordnung zu den jeweiligen Muskelsystemen war.

## 5 Ausblick

Die vorliegenden Ergebnisse widersprechen einer singulären, den Rumpfmuskeln per definitionem zuordenbaren Funktionsweise.

Die international eingeführte und auch allgemein als verbindlich angesehene Einteilung in die definierten Systeme (Bergmark 1989, Comerford und Mottram 2001) kann nach den vorliegenden Ergebnissen so nicht aufrecht erhalten werden. Die postulierte Funktionsspezifik wurde für jeden der untersuchten Rumpfmuskeln durchbrochen. Vielmehr kann jeder dieser Muskeln sowohl stabilisierende und auch mobilisierende Aufgaben wahrnehmen.

Für die Sicherung des Equilibriums aus Mobilität und Stabilität spielt offensichtlich also das fein abgestimmte Zusammenspiel aller Rumpfmuskeln eine entscheidende Rolle (McGill et al. 2003, McGill 2001). Dabei werden die Muskeln in ganz unterschiedlicher Art und Weise eingesetzt. Bisher gab es für alle vorgestellten Anwendungen keine Normwerte. Diese liegen nunmehr vor und können für weiter gehende diagnostische Fragestellungen verwendet werden. Insbesondere Patienten mit CURS bilden nach Ansicht des Autors hierbei die Hauptzielgruppe. Bei diesen Patienten verfehlt bisher ein Großteil der angewandten diagnostischen Maßnahmen ihr Ziel (Phillips et al. 1986, van Tulder et al. 1997). Das Krankheitsbild wird nach wie vor als Ausschlussdiagnose definiert (Müller 2001).

Die in der Literatur bereits vorliegenden funktionellen Befunde, die neben der verminderten Kraftausdauerfähigkeit von Patienten mit CURS vor allem auch Alterationen der normalen Rumpfmuskelkoordination identifizieren konnten, stehen nur scheinbar im Widerspruch zueinander. Erfolgreiche Therapiekonzepte des CURS setzen multimodale Behandlungsstrategien ein (Hildebrandt et al. 1997). In einem derartigen Therapiekonzept werden neben der ebenfalls notwendigen psychologischen Betreuung der Patienten sowohl kräftigende als auch koordinationsschulende Maßnahmen eingesetzt. Damit profitieren die behandelten Patienten am meisten, unabhängig davon, auf welcher Störungsebene ihr Problem primär angesiedelt ist.

Die vorgestellten Ergebnisse beinhalten diagnostische Aussagen während Alltagstätigkeiten, applizierten zyklischen Provokationen und statischen Tests der Rumpfmuskeln. Aus Sicht des Autors ergibt sich damit die Möglichkeit, spezifische Störungsebenen zu identifizieren. Eine mehrdimensionale Diagnostik des motorischen Systems, das sowohl statische als auch zyklische Provokationen umfasst wäre somit in der Lage, bestehende Defizite von CURS-Patienten genauer zu kennzeichnen. Somit könnte die Therapie von Patienten mit CURS effizienter gestaltet werden.

Die Einteilung der Rumpfmuskeln in die etablierte Systematik lässt weiterhin die nachweisbaren systematischen lokalisationsabhängigen Unterschiede zwischen ventralen und dorsalen Rumpfmuskeln unberücksichtigt.

Sowohl während dynamischer als auch während statischer Beanspruchung konnten in der vorliegenden Arbeit lokalisationsbezogene Spezifika der untersuchten Rumpfmuskeln herausgearbeitet werden.

Die histomorphologische Ausstattung der untersuchten Muskeln beeinflusst offensichtlich deren Reaktionsweise deutlicher als bisher angenommen.

Die muskuläre Antwort unterliegt weniger stark der Intensität der Beanspruchung, sondern wird offensichtlich auch durch die Richtung des einwirkenden Kraftvektors bestimmt.

Alle untersuchten Rumpfmuskeln können während dynamischer Anforderungen sowohl Bewegungen initiieren (konzentrische Kontraktion) als auch deren Ausführung kontrollieren und begrenzen (exzentrische Kontraktion).

Die Umsetzung der gefundenen Resultate in Diagnostik und vor allem auch in therapeutische Überlegungen bedeuten als Konsequenz die Abkehr von einer einseitig zuzuordnenden Funktionscharakteristik der untersuchten Rumpfmuskeln. Vielmehr sollte deren Funktionsheterogenität in Abhängigkeit von den jeweils wirkenden Anforderungen berücksichtigt werden.

Somit sind für Patienten mit CURS sowohl dynamische als auch statische Übungen, immer verbunden mit koordinativen Anforderungen zu empfehlen. Inwieweit einzelne Bereiche dieser Funktionalität beeinträchtigt sind, kann mit Hilfe der Normwerte aus den vorgestellten Studien im Detail diagnostiziert werden.
Denkbar ist weiterhin, die Normwerte auch im Rahmen präventiver Ansätze zu verwenden um frühzeitig funktionelle Defizite zu identifizieren und somit gezielt beeinflussen zu können.

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## 7 Originalarbeit 1

### Trunk muscle activation patterns during walking at different speeds

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**Abstract**

Investigations of trunk muscle activation during gait are rare in the literature. As yet, the small body of literature on trunk muscle activation during gait does not include any systematic study on the influence of walking speed. Therefore, the aim of this study was to analyze trunk muscle activation patterns at different walking speeds. Fifteen healthy men were investigated during walking on a treadmill at speeds of 2, 3, 4, 5 and 6 km/h. Five trunk muscles were investigated using surface EMG (SEMG). Data were time normalized according to stride time and grand averaged SEMG curves were calculated. From these data stride characteristics were extracted: mean SEMG amplitude, minimum SEMG level and the variation coefficient (VC) over the stride period. With increasing walking speed, muscle activation patterns remained similar in terms of phase dependent activation during stride, but mean amplitudes increased generally. Phasic activation, indicated by VC, increased also, but remained almost unchanged for the back muscles (lumbar multifidus and erector spinae) between 4 and 6 km/h. During stride, minimum amplitude reached a minimum at 4 km/h for the back muscles, but for internal oblique muscle it decreased continuously from 2 to 6 km/h. Cumulative sidewise activation of all investigated muscles reached maximum amplitude during the contralateral heel strike and propulsion phases. The observed changes argue for a speed dependent modulation of activation of trunk muscles within the investigated range of walking speeds prior to strictly maintaining certain activation characteristics for all walking speeds.

## Introduction

Bipedal walking is the most common means of locomotion for humans. This is unique among mammals. Occasionally, apes and other mammals solely use their legs for locomotion, but only intermittently between using all four extremities. Human gait has been investigated extensively. Starting with Eadweard Muybridge (1830-1904) and Etienne-Jules Marey (1830 - 1904) the main topic of investigation in movement examinations has been the analysis and modeling of human gait [1,9,33,41,48]. Functional parameters of muscles in human [10,19,44] as well as in animal gait [3,4,11,15,43] are analyzed to develop models of how walking is organized in general [5,35].

Although leg and hip muscles are the main walking actuators, the whole body is involved in locomotion. Opposite arm swings and rotational movements of the trunk [18] are also typical attributes of walking activities. Throughout all of these complex activities, both, stability and mobility of the vertebral column are realized. Together with joints and ligaments, our back muscles and trunk muscles assure the flexibility and integrity of the spine. This capacity is effective through an ensemble of differentially organized muscles. Large powerful muscles like the erector spinae muscle chiefly execute movements. Stability, on the other hand, does not necessarily require high muscle forces [31].

Trunk muscles have been divided into two muscle systems [2]: the local system ensuring stability and the global system enabling movements. There are two distinct types of activation patterns: Local system muscles are permanently active at low levels [8], independent of movements. Conversely, muscles of the global system act to initiate movements leading to movement dependent phasic activation patterns [2,8]. Recently, the global system was subdivided further into the global stabilizing and the global mobilizing systems [8,17]. Global stabilizers complement the function of the local system by controlling and limiting movements by means of eccentric activation characteristic [8]. Global mobilizers initiate movements [8].

Trunk muscles have all been assigned to one of these muscle systems. There is a paucity of functional investigations to support the assignment of trunk muscles to either the global or local system or to adequately characterize normal trunk muscle activation patterns during specific tasks. When such investigations have been carried out, for instance for the lumbar multifidus and the transversus abdominis muscles [25,26], it enabled the discovery, in further investigations, that patients suffering from chronic unspecific low back pain showed disturbed morphology [16,21,22] and functionality [27,28].

Functional investigations, systematically focusing on the activation characteristics of trunk muscles during gait are rare [6,32,42]. Typically in such studies needle EMG was used [32,42] making a transfer to routine diagnostic applications difficult. The aim of this investigation was to identify muscle activation patterns of trunk muscles during treadmill walking as a function of walking speed to establish normative data. These data can be used for identification of corrupted muscle co-ordination, especially in the diagnosis of low back pain patients

**Methods**

Fifteen healthy men (24-35 years, median: 27 years, for details of demographic data see table 1) with no history of low back pain voluntarily participated in this study. The study was approved by the local ethics committee of the University of Jena (0558-11/00) and, therefore, fulfils the declaration of Helsinki. Informed written consent was obtained from each volunteer. After an adequate habituation phase of five minutes, every subject walked on a treadmill (width: 62 cm, length: 160 cm) for approximately one minute at each of the randomly assigned speeds of 2, 3, 4, 5 and 6 km/h (0.56/0.83/1.11/1.39/1.67 m/s), respectively. Volunteers walked barefoot with normal arm swing. Bipolar surface EMG (SEMG, 5 Hz – 700 Hz, Biovision, Wehrheim, Germany) was taken from five pairs of trunk muscles [20,38], electrode positions see table 2. Disposable Ag-AgCl electrodes (H93 Arbo®, Germany) with a circular uptake area of 1 cm diameter and an inter electrode distance of 2.5 cm were used. Simultaneously, force data from both heels and pads were measured quantitatively to identify stride phases in detail (i.e. heel contact, pad contact, propulsion phase, and toe off). Data were stored for further offline analysis (AD-conversion at 2000/s, DAQCard-AI-16E-4: 12 bit, National Instruments, USA). Stride cycles (Cadence, left heel strike to left heel strike in this investigation) were identified with a semi-automatic software algorithm using the force signals from both heels, including visual control.

Table 1: Demographic data of investigated healthy subjects

| | age [years] | height [cm] | weight [kg] | BMI |
|---|---|---|---|---|
| Mean | 26.9 | 180.9 | 76.7 | 23.5 |
| SD | 3.0 | 6.5 | 7.1 | 2.0 |

BMI: body mass index

SD: standard deviation

Cadence time was analyzed and only strides within 25% deviation from the calculated median time of all respective strides were used for analysis. The number of strides used for calculation per subject varied from 30 to 80, depending on the exact recording time, the treadmill speed and on the number of eliminated strides due to technical problems. Raw SEMG was centered and high-pass filtered ($4^{th}$ order Butterworth filter, 20 Hz) to avoid influences from movement artifacts. A root mean square (RMS) envelope was calculated subsequently using a smoothing window of 50 ms. The data per stride were averaged after time-normalization to avoid variances originating from remaining variability in stride length. Time normalization had an accuracy of 0.5 % (201 data points). Grand averaged SEMG curves were calculated for all applied velocities, muscles, and subjects, respectively. Additionally, individual cumulative sidewise activation of all investigated muscles was determined by adding up the RMS values of all five different muscles of one side.

From the filtered and rectified curves, parameters characterizing whole strides were calculated: mean amplitude, minimum amplitude and variation coefficient (VC, calculation: SD/Mean*100%) of all calculated amplitude values during a stride.

Friedman one way ANOVA by ranks (SPSS®) for dependent samples was used to test speed dependent differences of the calculated time independent parameters.

Table 2: Investigated trunk muscles and respective electrode positions (according to [20,37]). Muscles from both sides were investigated simultaneously

| Muscle | electrode position and orientation |
|---|---|
| M. rectus abdominis (upper part, RA l/r) | 4 cm lateral navel, lower electrode at navel level, vertical |
| M. obliquus internus abdominis (OI l/r) | along horizontal line between both ASIS's, medial from inguinal ligament |
| M. obliquus externus abdominis (OE l/r) | upper electrode directly below most inferior point of costal margin, on line to opposite pubic tubercle |
| M. multifidus (lumbalis, MF l/r) | 1 cm medial from line between PSIS's and 1$^{st}$ palpable spinous process, lower electrode at L4 level, parallel to line |
| M. erector spinae (longissimus, ES l/r) | over palpable bulge of muscle (approx. 3 cm lateral midline) lower electrode at L1 level, vertical |

ASIS: anterior superior iliac spine
PSIS: posterior superior iliac spine

**Results**

For the investigated muscles grand averaged SEMG curves changed shape differently (figure 1). The small and constant low amplitudes for RA at the lowest velocities increased for speeds higher than 4 km/h and became more phasic (av. rank of VC, left RA 2-6 km/h: 1.3/2.5/3.1/3.7/4.4, see also figure 2). Peaks occurred at ipsilateral heel strike and at ipsilateral as well as contralateral propulsion, but amplitudes remained at comparably low levels. Generally, grand averaged curves of all other muscles did not change their main characteristics. For OI and OE highest amplitude peaks were identified during contralateral propulsion phase. Smaller but distinct amplitude peaks for OI coincided with ipsilateral, for OE with contralateral heel and pad contact, respectively. Peak amplitudes increased

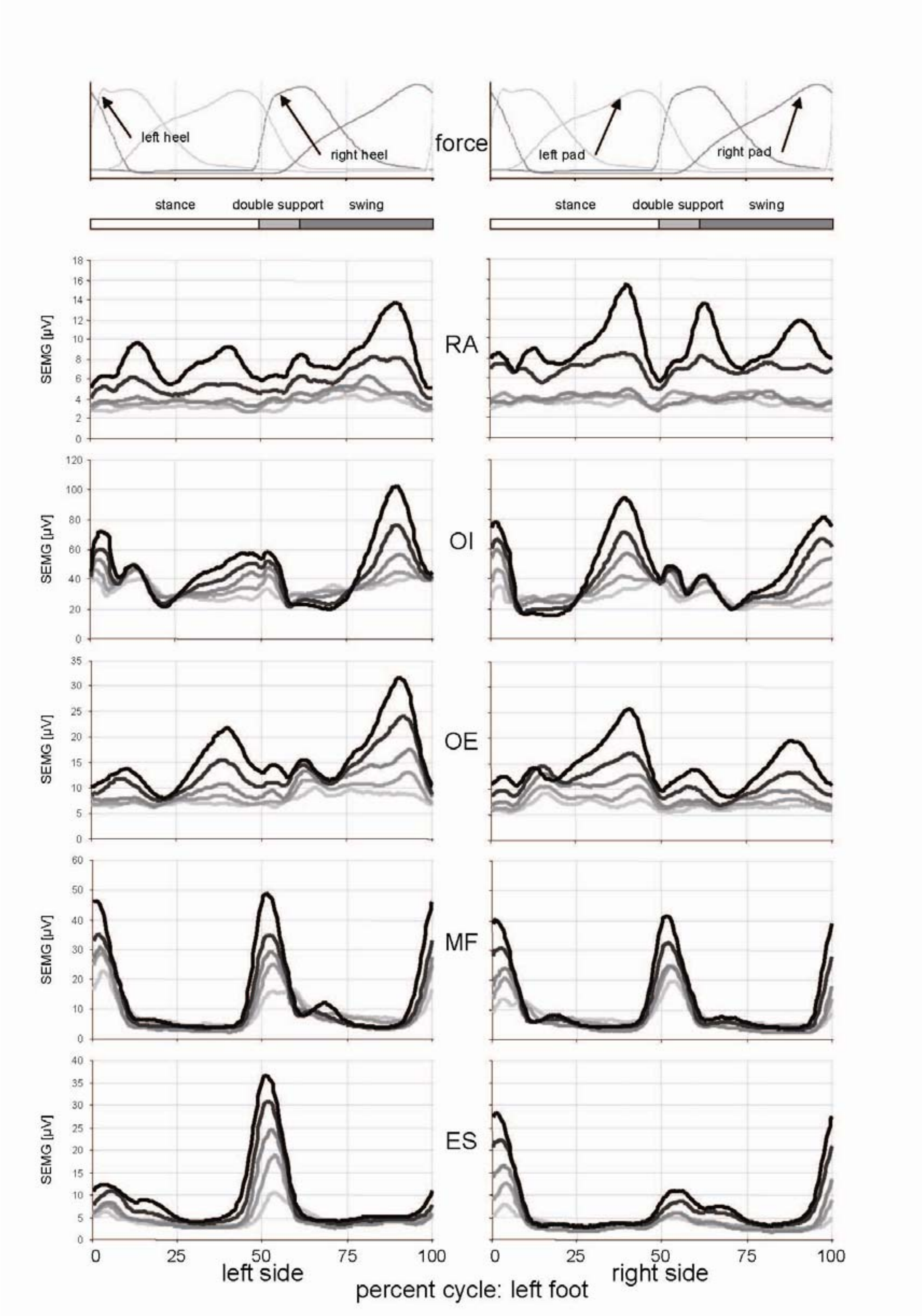


figure 1: Grand averaged SEMG curves of all investigated trunk muscles for the applied treadmill speeds of 2, 3, 4, 5 and 6 km/h. Colors gradually get darker with increasing speeds, therefore light grey represents 2 km/h, black represents 6 km/h. In the upper row force signals of heels and pads are displayed. Curves are time normalized for left stride

with increasing speeds. But for OI increasing treadmill speeds were accompanied by lowered minimum levels (av. rank of minimum amplitude, left OI 2-6km/h: 3.9/3.3/3.1/2.5/2.1). Mean and minimum amplitudes both increased with increasing speeds for OE, but VC level increased also, indicating larger amplitude ranges. This mainly originated from the more distinct increase in maximum amplitudes.

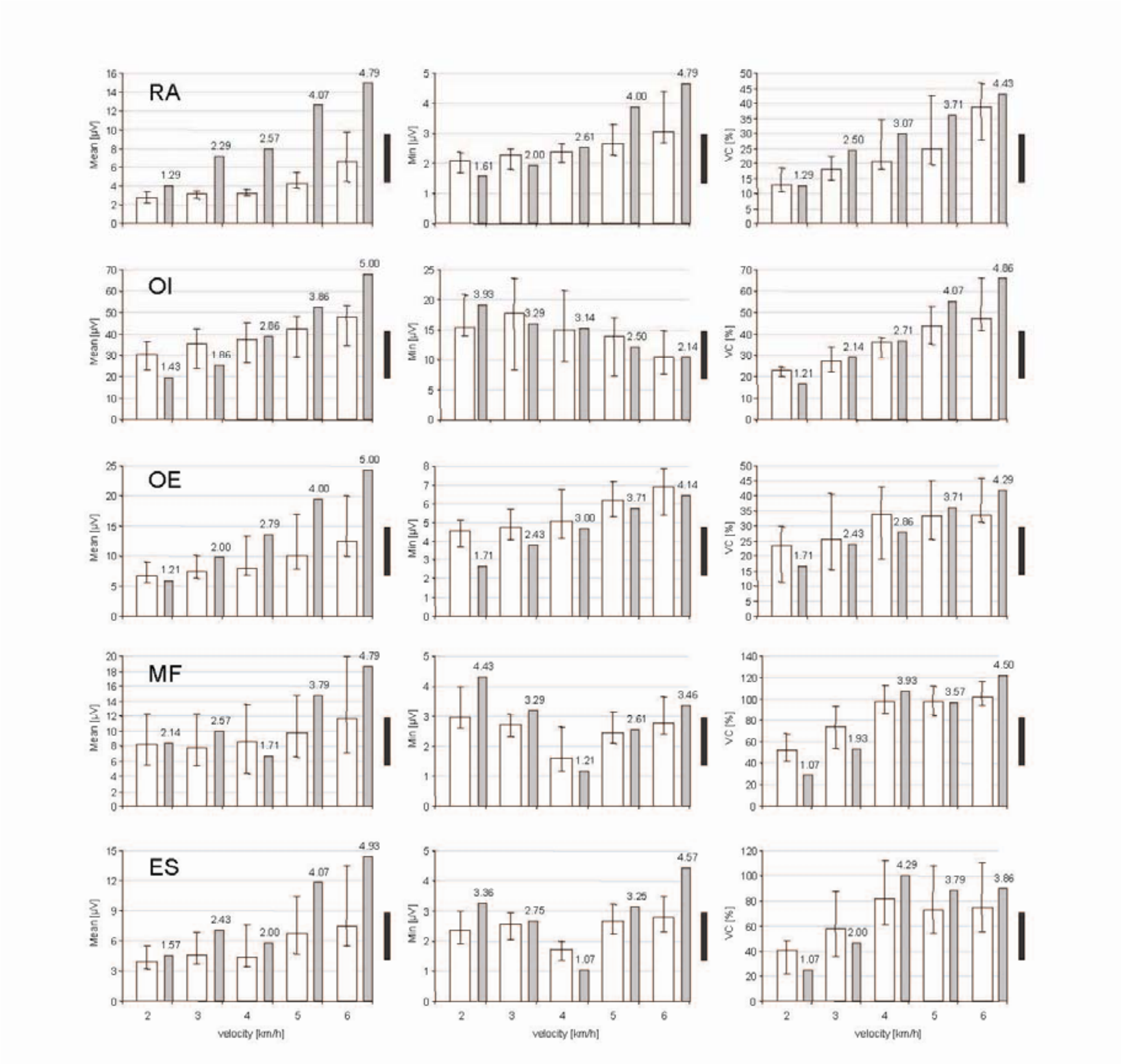


figure 2: Time independent parameters during stride for all applied treadmill speeds. Left column: mean amplitude, middle column: minimum amplitude, right column: VC. Grey bars and numbers indicate mean rank values from the friedman test. Dark rectangles indicate the critical rank difference (1.60) for at least p<0.05 significance level.

Both, MF and ES muscles were characterized by increases of their amplitude peaks during heel strike, but their minimum amplitudes did not change much. MF and ES mean amplitude remained virtually unchanged up to 4 km/h, but increased at higher speeds. Minimum amplitude decreased from 2 to 4 km/h and increased at 5 and 6 km/h (see figure 2). Minimum amplitude for MF was highest at 2 km/h (av. rank 4.4), for ES at 6 km/h (av. rank 4.6). VC showed virtually unchanged levels from 4 to 6 km/h in both muscles (figure 2).

The cumulative amplitude of all investigated trunk muscles of one side reflects general speed dependent activation characteristics: activation peaks at ipsilateral heel strike and pad contact as well as during contralateral heel strike and propulsion phase increased with increasing speed. In contrast, low-level activations during stance phases remained virtually unchanged and therefore independent from walking speed (figure 3).

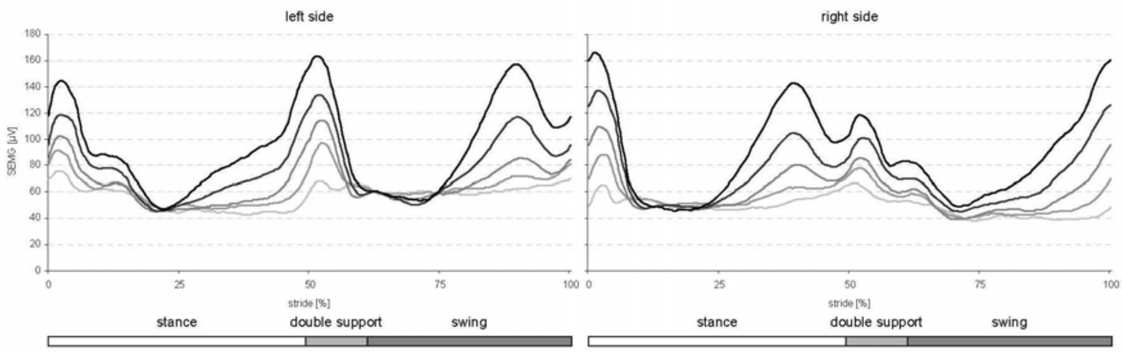


figure 3: Cumulative amplitude of all investigated trunk muscles of one side, displayed as grand averaged curve for treadmill speeds of 2, 3, 4, 5 and 6 km/h. Colors gradually get darker with increasing speeds, therefore light grey represents 2 km/h, black represents 6 km/h. Curves are time normalized for left stride.

## Discussion

Initially, the assignment of trunk muscles to either the local or global muscle systems was based on mechanical analysis, but function remained hypothetical [2]. More detailed understanding proceeded from functional investigations of transversus abdominis [24,26,30] and morphometric data for lumbar multifidus muscles [22]. Since stability in vertebral neutral position can only be maintained by active (muscular) components [39], further evaluation of muscle function seems crucial for understanding vertebral stability and the development of low back pain. In this context the complex system of trunk muscle co-ordination plays a major part [34,40,46].

OE activation pattern, although showing strong similarity with OI pattern is characterized by distinct differences: preparation (i.e. ipsilateral propulsion phase) for contralateral heel strike is more pronounced in OE and its minimum amplitude levels, together with VC, increase with increasing speed throughout the whole investigated speed range.

In general, investigated back muscles were characterized by similar speed dependent changes: Mean amplitudes remained consistently low up to the 4 km/h mark, then they started to increase at 5 km/h. Lowest minimum amplitude levels were observed at 4 km/h, accompanied by highest VC levels. However, these comparable characteristics of time independent data do not highlight the principle differences in activation patterns of both back muscles. MF activation was characterized by two virtually equal activation peaks during both heel strikes, whereas ES only showed one relevant peak during contralateral heel strike. Therefore, activation patterns during stride differ explicitly, but speed dependent alterations generally follow the same rules.

According to convention, all investigated muscles have been assigned to either the local or global muscle systems [2,8]: RA and ES are global mobilizers, whereas OI and OE belong to the global stabilizers. Deep fibers of MF are essential parts of the local system, but the superficial fibers act more phasically, and therefore in functional analogy with global stabilizing muscles [36]. In this study the expected activation characteristics did not consistently support this differentiation. At low speeds trunk movements are only slight, but become larger and faster with increasing speeds. Therefore, the equilibrium of stability and mobility demands changed more rapidly at higher speeds and muscles changed activation characteristics according to these changing functional demands. Observed amplitude peaks for OE and MF during contralateral heel strike indicate eccentric activity. Concentric activation characteristics were evident as well: for MF, while producing amplitude peaks during ipsilateral heel strike, for OE during ipsilateral heel strike, pad contact and ipsilateral propulsion phase, and for OI during contralateral propulsion. ES activation during contralateral heel strike most probably was eccentric again, because of the movement of the spine and the load transfer in the lumbar region [47].

Therefore except RA, which was not distinctly involved during walking, all other investigated trunk muscles were characterized by speed dependent changes of their activation characteristics. OI and OE muscles, starting with continuous activation at low speeds developed mixed phasic patterns reflecting both, global mobilizing and stabilizing characteristics [8]. MF muscle, almost inactive during slow walking speeds, increased amplitude peaks at heel contacts, again reflecting mobilizing and stabilizing characteristics. ES eccentric activation matches characteristics related to global stabilizing muscles.

For MF bipolar SEMG is only able to represent superficial activation patterns [7], which have been proven to differ from deep activities, at least during preparatory activations prior to arm movements [36].

Although electrodes were positioned carefully by one of two experienced examiners according to international recommendations [20,37], cross talk could not be avoided completely [12,14]. This is a general problem in surface EMG [13].

A comparable problem of different muscle depths arises at least for OI: similar fiber directions at ASIS height argue for cross talk at least from transverse abdominal muscle. However, the possibility of interfering representation of both, internal oblique and transverse abdominal muscles in the measured SEMG gives one a clue about the stability situation of the ventral trunk wall, although exact proportions of these two muscles remain unknown in the actual investigation. The direct neighborhood between MF and ES also necessarily causes cross talk from ES for MF measurements [45], but the different patterns between both muscles indicate at least partial selectivity of the measured SEMG signals.

The changed pattern of OI from continuous activation at low speeds into phasic activation at higher walking speeds, indicated by the highest VC values at 6 km/h, may be due to an increased proportion of OI activation within the interference signal. Transverse abdominal muscle, which has been investigated extensively [23,26,29] and is assigned to the local muscle system [8], should not contribute to this. The decreasing minimum amplitude levels at higher speeds could indicate a changed activation pattern for transverse muscle too, but rather may be due to low cross talk interference [42] and therefore decreased OI activity.

Independent from strategies of individual muscles during the various walking speeds, activation characteristics can be evaluated in terms of the cumulative amplitude of all investigated trunk muscles of one side of the body (figure 3). In this way, an heightened response could be identified during increased speeds. Specifically, amplitudes increased with walking speed during heel strike and propulsion. After pad contact overall activity remained virtually unchanged at all speeds. Therefore, heel strike as well as propulsion require general speed dependent adaptations of trunk muscle activation. The differently organized trunk muscle activation patterns during the varying walking speeds result in a consistent increase of overall activity. This argues for the necessity of compensation of the higher rotational momentum forces but also for increased stabilization demands [34], since trunk activity during gait is not essential for the walking process itself.

## Conclusions

Trunk muscle amplitude levels increase but activation patterns change differently with increasing walking speeds. For OI muscle a change from continuous towards phasic activation pattern could be observed. MF and ES muscles are both characterized by clear phasic activation patterns. The results therefore argue for task i.e. walking velocity dependent change of muscle function characteristics instead of strictly maintained function within the assigned muscle system.

## Acknowledgement

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# 8 Originalarbeit 2

## Activation characteristics of trunk muscles during cyclic upper body perturbations caused by an oscillating pole

Christoph Anders, Beatrix Wenzel, Hans-Christoph Scholle

**Archives of Physical Medicine and Rehabilitation, 2008;89(7):1314-1322**

**Abstract**

**Objective:** Back problems are often related to disturbed muscular coordination. There is still a need for therapies that target key muscles to establish or restore normal functional characteristics. This investigation was carried out to evaluate the effect of a new device on trunk muscle activation.

**Design:** Crosssectional survey of trunk muscle activation characteristics.

**Setting:** Physiological laboratory at university institute.

**Participants:** 30 healthy subjects (15 women, 15 men) were recruited from a university campus.

**Interventions:** A simple flexible pole that applies rapidly alternating forces on the trunk when set into motion was used. The device was held in both hands, in front of the body. It was used at three different oscillation frequencies (3 Hz, 3.5 Hz, 4.5 Hz), in horizontal and vertical plane, respectively.

**Main Outcome measures:** Surface EMG of five trunk muscles was measured and the data were normalized according to relative cycle time. Time dependent (amplitude curve) and time independent (mean amplitude over cycle) parameters were used for analysis.

**Results:** Rectus abdominis and external oblique muscle amplitudes were directly proportional with oscillation frequency (ANOVA), and these effects were independent of gender. Multifidus amplitude levels were subject to oscillation plane with increased levels for vertical oscillation in men but not in the women's group.

All abdominal muscles exhibited continuous activation pattern, independent of oscillation plane. Back muscles changed from a continuous activation in horizontal

plane into similarly phasic patterns in vertical oscillation plane. The occurring amplitude peak moved forward in relative cycle with increasing oscillation frequency.

**Conclusions:** Back muscle activation patterns were subject to oscillation plane. Abdominal muscle activation was independent from oscillation frequency and oscillation plane. These normative data may be used to identify disturbed trunk muscle coordination patterns and to control success of functional restoration during rehabilitation interventions of back pain patients.

## Introduction

Localized lower back pain is often related to muscular problems. This is the main message conveyed in summarizing the physiological research of chronic non-specific low back pain (LBP) during the last two decades [1-8].

Two different kinds of disturbances in patients with LBP have been identified: reduced endurance of their back extensors [9,10], and delayed essential feed forward postural responses of deep abdominal muscles [6].

Reduced endurance of back extensors may be due to chronic under-use of the respective muscles due to a lack of appropriate physical activity. At present the question remains unanswered as to whether the pain is generally a result of physical deconditioning or vice versa. Since pain usually reduces physical activity it is hard for these patients to get out of the vicious circle of pain, leading to reduced mobility, which leads to even more reduced physical activity. Either way, intense physical training of any kind [11] is often able to reduce pain, but long term success remains elusive if training is not maintained [12,13].

Delayed postural responses in trunk muscles [6] may lead to impaired coordination of trunk muscles in patients with LBP. One possible reason for this might be a cumulative effect of microlesions of proprioceptors at the spine [14]. This hypothesis is supported by other results that prove impaired movement perception of both Patients with LBP [15] and patients with lumbar stenosis [16].

The identification of disturbed coordination patterns led to the development of specific training programs targeting deep abdominal as well as back muscles [17]. One of these programs was able to reduce LBP, lasting for a critical observation period of two years [18]. This spinal segmental stabilization program involves a combination of biofeedback guided voluntary activation of the deep transverse abdominal muscle and therapist-assisted voluntary activation of the multifidus muscle.[17]. The basic principle for this training originates from the assignment of all trunk muscles to either the local or the global muscle systems [19]. These muscle systems are believed to behave in a predictable manner defined by their function: Local muscles are thought to constantly remain active at low levels, providing segmental stability; global muscles should activate in a phasic manner, to initiate or limit movements [20].

Results from our own studies could not completely confirm these different classifications. For the back muscles, no substantial differences of fiber type distribution patterns have been identified between mobilizing (erector spinae muscle) and stabilizing (multifidus muscle) muscles [21]. Perhaps this explains why during static tasks both muscles acted in an identical manner [22]. It does not suggest, however, why activation characteristics during dynamic requirements (treadmill walking) were different, with eccentric and concentric activations for the multifidus muscle and more pronounced eccentric activations for the erector spinae muscle [23]. Analogous findings during static tasks could be identified for abdominal muscles: The amplitude-force relationship of several abdominal muscles was virtually identical [22], independently from whether they were assigned to the mobilizing (rectus abdominis muscle) or the stabilizing muscle systems(both oblique abdominal muscles).

Overall these results seem to conflict, but from our point of view, they simply indicate different aspects of functional pathology, which can be found to varying degrees among the Patients with LBP. Patients with LBP most probably are characterized by inadequate physical condition and impaired trunk muscle coordination. Previously it has not been possible to definitely diagnose the degree of either one.

To summarize, patients with LBP exhibit two kinds of muscular disturbances: inadequate force capacity, supposedly caused by chronic under use, and disturbed coordination, most probably due to impaired proprioception. For the first problem, any kind of physical training can improve status short term, but for long term effects adequate muscular coordination is probably a critical factor.

The Spinal Segmental Stabilization Program [17] involves high labor costs because it requires a two-step training process: the initial strengthening and training and then a subsequent training, because the voluntary activation of the trained muscles has to be transferred back into involuntary muscle function for performing everyday activities. Therefore, there is still a need for therapeutic techniques that activate the target muscles involuntarily and are robust against external influences. A better solution would incorporate techniques that automatically train the stabilizing muscles. The authors received notice of a new device that can roughly be described as a simple flexible pole. This pole can be set into oscillation, resulting in tunable, rapidly alternating forces. By holding this device in both hands in front of the body, forces act on the whole trunk. Hypothetically, this should trigger activity of all trunk muscles.

The study was carried out to determine the effect of using this device on trunk muscle activity and, furthermore, to determine the influence of changes in oscillation frequency and oscillation plane of the device.

Hypotheses

Local muscles are expected to act at continuous activation level (tonic activation), mostly independent from oscillation plane and oscillation frequency. On the other hand, global muscles are expected to show alternating activation patterns (phasic activation) which should clearly be influenced by oscillation plane and oscillation frequency of the test device.

**Methods**

In this study 30 healthy subjects participated voluntarily. Prior to the investigation written informed consent was given by all persons. The study was part of larger experimental setup and was approved by the Jena University ethics board (0558-11/00). The group consisted of 15 women and 15 men (anthropometric data see table 1).

Table 1: Anthropometric data of the investigated subjects

| | | female | male | p |
|---|---|---|---|---|
| height | [cm] | 169.9 ± 5.8 | 179.7 ± 3.9 | 0.000 |
| weight | [kg] | 58.8 ± 5.1 | 72.7 ± 7.8 | 0.000 |
| BMI | | 20.4 ± 1.5 | 22.5 ± 2.5 | 0.008 |
| age | [years] | 23.1 ± 2.0 | 25.5 ± 5.7 | 0.154 |

Statistics: Student's Test for independent samples

values are displayed as mean ± SD

The exercise involved initiating and maintaining the oscillation of the device[a] (Propriomed®) pole ends with both hands held at the same height in front of the body (figure 1). The grip located in the middle of the device has a width of about 20 cm that enabled a secure hand hold for both hands. Attention was paid to correct task execution of the evoked oscillations with only one node, located in the middle of the pole (see figure 1). After oscillation initiation virtually no further movements of the handle were made, save for the small impulses to maintain the oscillations. The device has no active parts. Therefore, every movement of the device is result of the interaction between subject and device.

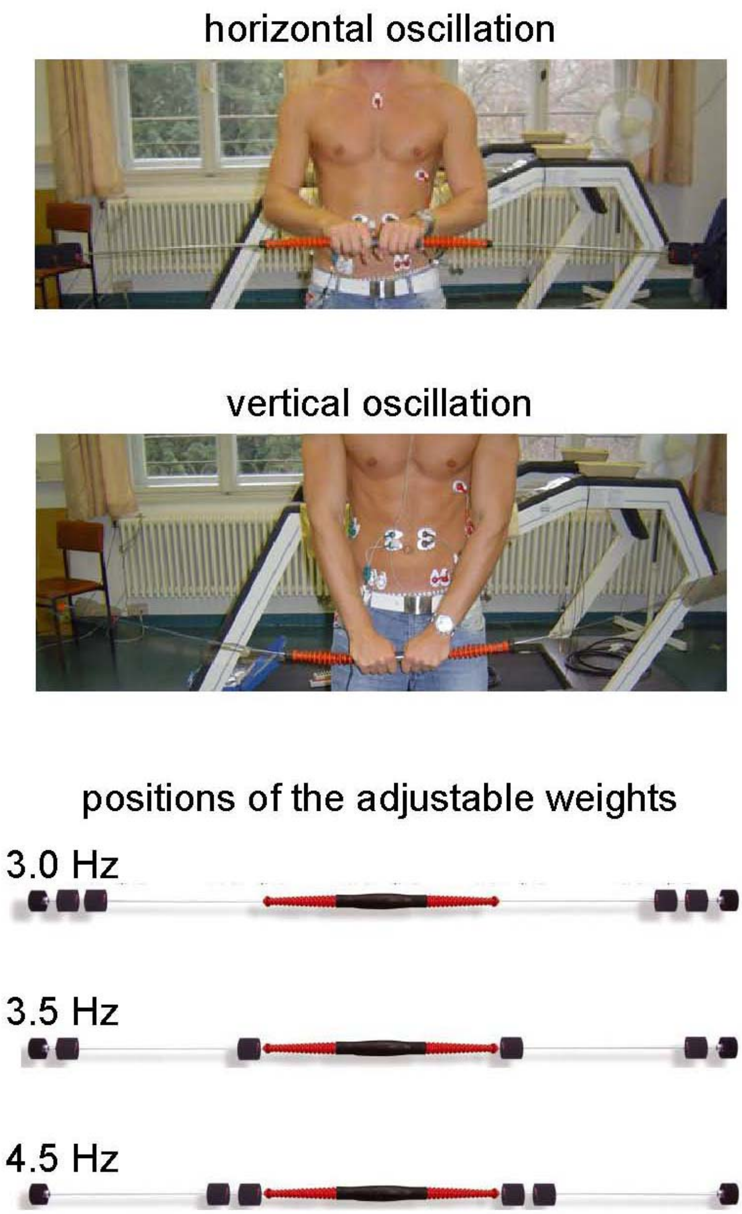


Figure 1: Subject using the device. Oscillation frequencies can be tuned by moving the adjustable weights. Weight positions and their respective oscillation frequencies are given.

The device had a total length of 170 cm and a weight of 1035 g, equipped with two adjustable weights at each side. The applied oscillation frequencies resulted from the positions of the adjustable weights: both positioned at most outside positions (3.0 Hz, slow), one positioned at the most inside position while the other remained at most outside position (3.5 Hz, moderate), and both being positioned at most inside positions (4.5 Hz, fast, see also Figure 1). The recommended oscillation amplitude of about 50 cm of the pole ends was maintained for all trials. Instead of a similar device (bodyblade) [24] which consists of a flexible foil blade the Propriomed® consists of a flexible rod, enabling oscillations in all possible directions of the sagittal plane. Therefore, in addition to maintain the requested oscillation amplitude, subjects had to pay attention to oscillation plane. Two oscillation planes were applied: horizontal plane and vertical plane. The order of the six possible plane-oscillation combinations was randomized separately for every volunteer. Each task was performed three times for each plane and frequency combination. For measurement, each trial lasted for at least 10s of stationary oscillation action. Sufficient breaks between the trials were maintained to prevent muscle fatigue (60 seconds between trial repetitions, 90 seconds between different tasks).

Table 2: Investigated muscles and electrode locations, according to [48,49]. Muscles from both sides were investigated simultaneously

| Muscle | electrode orientation and position |
|---|---|
| M. rectus abdominis (upper part, RA) | 4 cm lateral umbilicus, vertical, caudal electrode at level of umbilicus |
| M. obliquus internus abdominis (OI) | along horizontal line between both ASIS's, medial from inguinal ligament |
| M. obliquus externus abdominis (OE) | cranial electrode directly below most inferior point of costal margin on line to opposite pubic tubercle |
| M. multifidus (lumbalis, MF) | 1 cm medial from line between PSIS's and 1st palpable spinous process, caudal electrode at L4 level |
| M. erector spinae (longissimus, ES) | vertical, over palpable bulge of muscle (approx. 3 cm lateral midline) caudal electrode at L1 level |

ASIS: anterior superior iliac spine.
PSIS: posterior superior iliac spine.

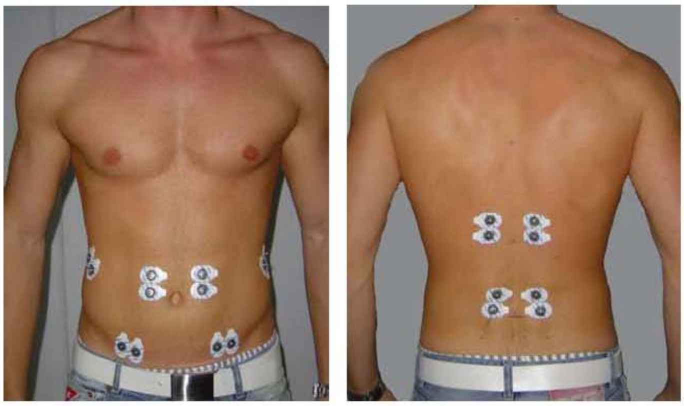

Figure 2: Electrode positions for all investigated trunk muscles. Detailed description for positioning is given in table 2. Positions were chosen according 48,49

Trunk muscle response was evaluated by Surface EMG (SEMG). Five different trunk muscles were investigated on both sides simultaneously (figure 2). The muscles and the respective electrode locations are detailed in table 2. SEMG was measured using a common bipolar montage with an inter-electrode distance of 2.5 cm. The circular uptake area of the disposable Ag-AgCl solid gel electrodes[b] had a diameter of 1 cm. Signals were amplified[c] with a gain of 2500 (-3dB between 5 Hz and 700 Hz). AD conversion was done at 2000 samples per second with an accuracy of 1 µV/bit (DAQCard -AI-16E-4[d]: 12 bit). A bidirectional acceleration[a] sensor was fixed on the right end of the pole to enable identification of correct oscillations with respect to plane. Only oscillation cycles which could be identified to match the required plane were used for further analysis (criterion: maximum time difference of 25 ms between both peaks). Therefore, the mean proportion of cycles for analysis varied between 78% (horizontal plane, fast) and 94% (vertical plane, moderate) of all performed cycles. Furthermore, only cycles with a temporal deviation of less than 10% from the median cycle time of every single trial were used for calculation. Raw SEMG signals were band pass filtered between 20 Hz and 300 Hz and a moving root mean square (rms, window: 15 ms) was calculated. To reduce slight differences in cycle duration and to compare the different oscillation frequencies all cycles were time normalized with an accuracy of 0.5% (201 time points). The normalized cycle starts at the rear inversion point (vertical oscillation: top inversion point), at 50% the front inversion point is reached (vertical oscillation: bottom inversion point).

For each subject a mean SEMG curve of each trial was calculated, separately for all muscles. A grand average was calculated by averaging the three respective single trials per plane and oscillation frequency. From these grand averaged data, time independent parameters (mean amplitude level) and time dependent parameters (muscle coordination patterns during oscillation cycle) were used for analysis. To identify coordination patterns, amplitude patterns were normalized as well according the maximum amplitude within cycle.

All statistical calculations were performed using the SPSS® package[e]. To test interactions of oscillation plane, oscillation frequency and gender for mean SEMG amplitudes a repeated measures ANOVA was used with gender as inter-subject factor.

Nonparametric tests were chosen for statistics of SEMG parameters. Influence of the two oscillation planes was tested using the Wilcoxon Test for dependent samples. Comparisons between the three oscillation frequencies were calculated using the Friedman one way ANOVA by ranks for dependent samples. Gender differences were tested using the Mann-Whitney Test for independent samples.

To test differences of the time dependent data, average values of sections of 10% of the normalized cycle were tested using the corresponding tests according to the time independent data.

## Results

### Time independent data

ANOVA revealed no interactions between the two intra-subject and the inter-subject factors (Table 3). Therefore, differences between genders, influence of oscillation frequency, and differences between oscillation planes could be calculated separately.

Significantly higher mean SEMG amplitudes were seen in the multifidus muscle (MF) in vertical plane when compared to horizontal plane at slow and moderate oscillation frequencies (table 4). Influence of oscillation frequency on mean SEMG amplitude level was only detectable in male subjects. SEMG amplitudes for external oblique (OE: both sides $p<0.05$) and internal oblique (OI: left: $p<0.05$, right: n.s.) muscles increased together with increasing oscillation frequency in vertical plane (table 4). Mean back muscle SEMG amplitudes increased with oscillation frequency in horizontal plane, significant for left MF and right erector spinae (ES) muscle. Also in horizontal plane, rectus abdominis (RA) muscle showed highest mean SEMG amplitude levels for moderate oscillation frequency (left: $p<0.05$, right: n.s.). Right OE showed highest mean SEMG amplitude levels for moderate oscillation frequency.

Table 3. P-values of the repeated measured ANOVA for mean amplitudes over normalized cycle. OI, and ES muscles did not show any significant difference. n.s.: not significant

| muscle | plane | plane * gender | frequency | frequency * gender | plane * frequency | plane * frequency * gender | gender |
|---|---|---|---|---|---|---|---|
| RA l | n.s. | n.s. | 0.018* | n.s. | n.s. | n.s. | n.s. |
| RA r | n.s. | n.s. | 0.019* | n.s. | n.s. | n.s. | n.s. |
| OE l | n.s. | n.s. | 0.024* | n.s. | n.s. | n.s. | 0.010* |
| OE r | n.s. | n.s. | 0.010* | n.s. | n.s. | n.s. | 0.038* |
| MF l | 0.010* | n.s. | n.s. | n.s. | n.s. | n.s. | 0.017* |
| MF r | 0.002** | n.s. | n.s. | n.s. | n.s. | n.s. | 0.017* |

* $p<0.05$

** $p<0.01$

Most mean MF SEMG amplitude levels were significantly higher in men than women (exceptions being fast frequency in vertical plane and left MF at slow and moderate frequency in horizontal plane). In contrast, women showed significantly higher OE amplitudes, except right sided OE for moderate and fast oscillation frequencies.'

See table 4 for mean SEMG data. Results of the statistical calculations are indicated.

Table 4: Mean SEMG amplitudes for all investigated trunk muscles. Values are displayed as median and upper and lower quartile range, respectively.

| muscle | | Horizontal | | | vertical | | |
|---|---|---|---|---|---|---|---|
| | | slow | moderate | fast | slow | moderate | fast |
| RA l | m | **5.0 (3.9/1.9)** | **6.5 (2.3/3.0)** | **5.6 (8.2/2.4)** | 3.9 (2.5/0.4) | 6.0 (3.4/1.9) | 4.8 (3.1/0.9) |
| | f | 4.0 (3.1/0.4) | 4.2 (5.1/0.7) | 4.4 (3.4/0.3) | 4.5 (2.6/0.8) | 5.4 (3.2/1.7) | 5.9 (1.3/1.1) |
| RA r | m | 4.9 (3.5/0.5) | 7.3 (2.1/3.0) | 5.0 (6.7/0.5) | 6.8 (1.2/1.7) | 7.0 (2.5/1.1) | 6.6 (2.6/0.6) |
| | f | 5.3 (2.4/0.5) | 5.9 (3.8/0.8) | 6.1 (3.1/1.2) | 5.4 (2.5/0.4) | 6.0 (3.9/0.4) | 6.7 (2.6/0.9) |
| OI l | m | 29.8 (8.1/17.1) | 28.7 (4.8/16.0) | 31.4 (15.3/19.7) | **28.0 (4.1/13.9)** | **29.3 (8.2/13.1)** | § **31.2 (6.8/11.8)** |
| | f | 22.4 (2.5/11.7) | 22.1 (2.8/9.9) | 21.0 (6.0/7.7) | 23.1 (2.3/11.1) | 21.3 (6.6/9.1) | § 20.1 (6.7/7.4) |
| OI r | m | 23.6 (11.2/9.8) | 25.4 (8.1/15.7) | 22.8 (16.2/13.6) | 23.7 (6.1/12.3) | 23.0 (12.7/11.5) | 28.1 (7.2/13.4) |
| | f | 13.5 (11.8/4.0) | 10.9 (16.9/1.5) | 12.5 (11.5/3.1) | 13.5 (9.6/4.5) | 14.3 (13.2/5.0) | 16.4 (12/6.4) |
| OE l | m | § 7.0 (1.7/1.5) | § 6.5 (5.0/0.9) | 7.3 (2.9/1.1) | § **7.2 (1.1/1.2)** | § **7.7 (1.8/1.4)** | § **7.9 (3.3/2.2)** |
| | f | § 11.2 (4.7/2.6) | § 11.1 (5.2/3.1) | 11.4 (8.0/3.0) | § 11.2 (3.3/2.3) | § 10.7 (6.2/2.4) | § 13.0 (6.8/5.0) |
| OE r | m | § **8.4 (3.3/2.0)** | **9.6 (3.2/3.6)** | **9.1 (8.0/1.9)** | § **7.3 (3.1/1.1)** | **9.2 (3.8/2.7)** | **9.4 (4.0/1.8)** |
| | f | § 11.0 (5.1/1.4) | 12.0 (8.5/2.6) | 12.5 (8.7/2.8) | § 12.3 (5.6/3.1) | 11.3 (7.9/2.7) | 11.8 (5.9/1.2) |
| MF l | m | **32.9 (11.9/15.1) *** | **31.0 (12.5/8.6)** | § **34.1 (14.4/11.2)** | § 36.6 (7.3/7.5) * | § 33.8 (10.7/5.1) | 33.6 (6.9/5.3) |
| | f | 19.8 (11.4/5.6) | 19.5 (7.4/4.5) | § 20.0 (13.7/6.6) | § 27.3 (3.6/11.2) | § 24.6 (8.3/6.6) | 23.5 (10.6/6.6) |
| MF r | m | § 30.5 (16.4/6.1) | § 32.5 (11.6/10.2) * | § 36.1 (11.1/12.0) | § 35.0 (12.2/7.5) | § 38.0 (7.5/9.2) * | 32.3 (10.7/5.3) |
| | f | § 20.5 (4.7/5.5) | § 19.1 (6.5/4.0) * | § 19.8 (6.6/7.3) * | § 21.0 (10.9/5.8) | § 20.4 (15.6/3.5) * | 19.4 (16.6/3.3) * |
| ES l | m | 26.9 (9.0/8.5) | 27.9 (8.3/9.2) | 28.0 (11.5/9.3) | 25.2 (11.3/5.4) | 27.3 (6.9/7.3) | 26.9 (6.4/10.8) |
| | f | 25.1 (14.4/9.4) | 27.0 (9.5/10.1) | 30.5 (7.9/10.9) | 22.1 (14.5/1.0) | 25.1 (7.9/5.0) | 25.5 (7.0/2.6) |
| ES r | m | **35.3 (3.9/15.4)** | **33.3 (7.4/11.8)** | **38.6 (3.1/19.4)** * | 27.9 (7.0/6.3) | 29.7 (9.1/4.4) | 29.8 (6.9/10.8) * |
| | f | 22.1 (12.2/3.6) | 24.5 (8.7/8.3) | 25.2 (15.4/7.4) | 25.8 (11.9/6.9) | 24.8 (11.0/5.1) | 28.5 (6.9/5.5) |

l: left side, r: right side

m: male, f: female

**bold**: significant differences between oscillation frequencies (Friedman ANOVA by ranks)

*: significant differences between oscillation planes (Wilcoxon)

§: significant differences between genders (Mann-Whitney)

### Time dependent data

Activation characteristics differed considerably between oscillation planes: horizontal oscillation evoked continuous activities in all muscles, whereas in vertical plane both back muscles exhibited clear phasic patterns (figure 3).

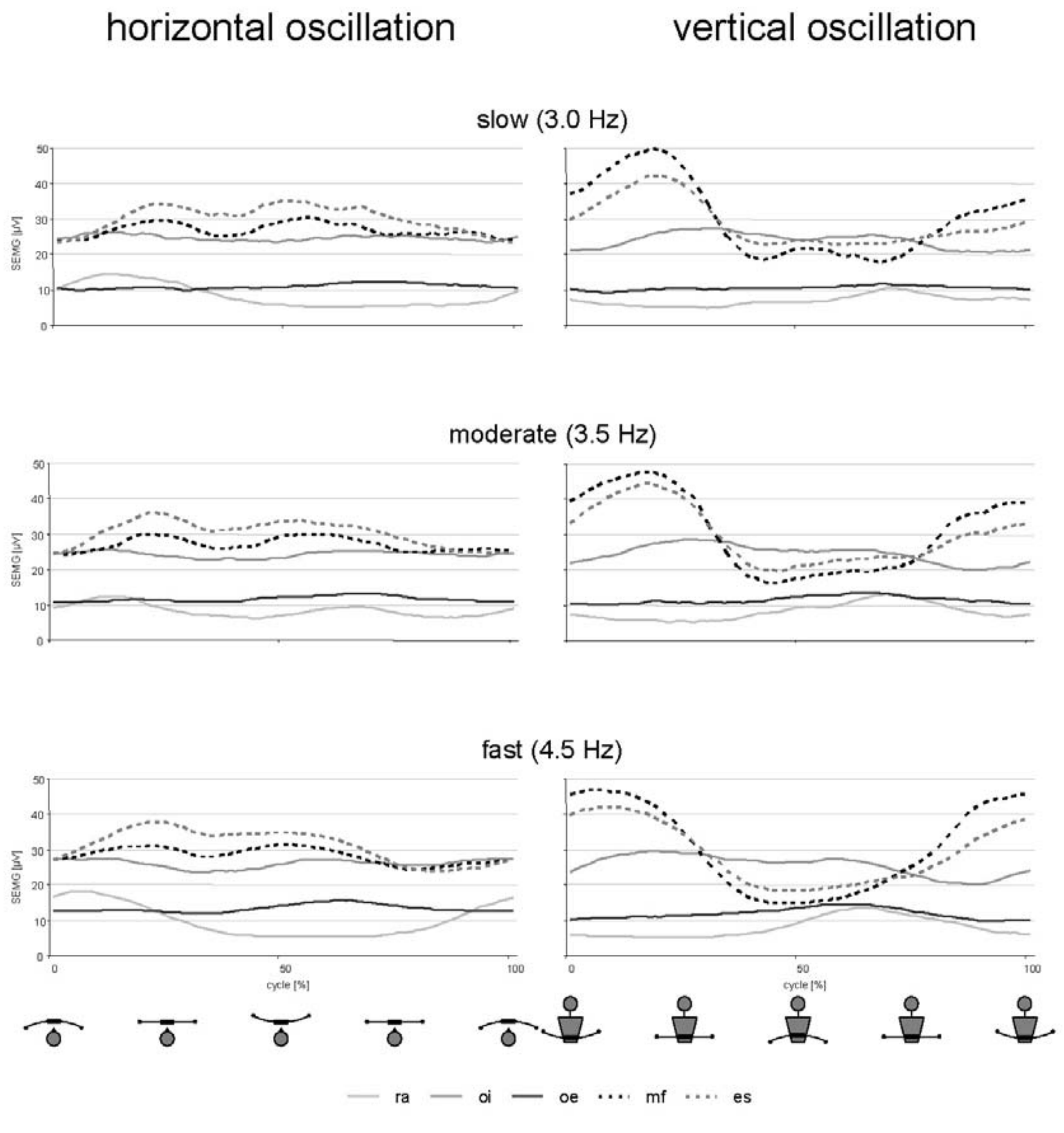


Figure 3: Grand average SEMG curves of trunk muscles during the applied oscillation frequencies in horizontal (left column) and vertical (right column) planes. Data from both genders were pooled. Positions of the pole ends during the normalized oscillation cycle are indicated.

Their highest amplitudes were reached right after top inversion point during downward movement of the ends of the pole. For the highest oscillation frequency this amplitude peak moved forward in terms of relative cycle time, more near the top inversion point (figure 4). Abdominal muscles, independent of oscillation plane, all exhibited continuous activation patterns which became only slightly phasic with increasing oscillation frequency. A clear order of amplitude levels could be identified: RA and OE with the lowest levels, followed by OI, MF, and ES in horizontal plane. Vertical oscillations exhibited slightly higher amplitude peaks of MF in comparison to ES, but minimum amplitudes were lower for MF.

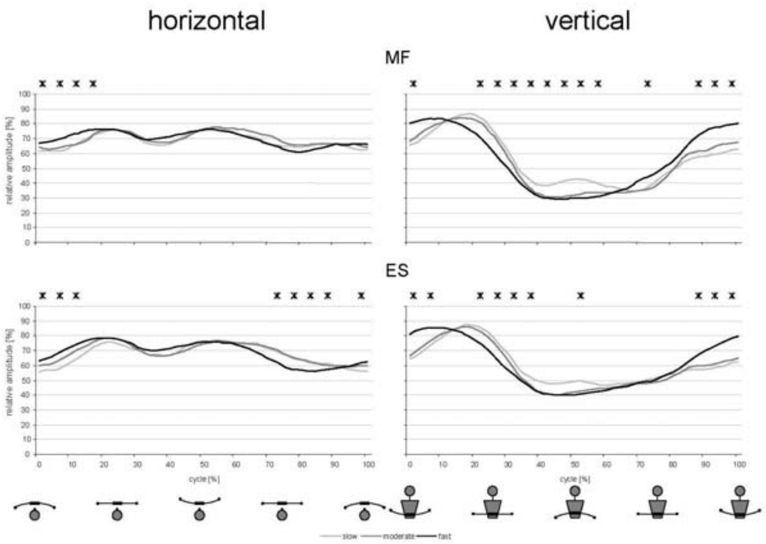


Figure 4: Coordination patterns of back muscles for oscillations in horizontal and vertical planes. Influence of oscillation frequency was tested by using the non-parametric Friedman ANOVA by ranks. Asterisks indicate significant differences of $p<0.05$. Data from both sides and genders were pooled. Positions of the pole ends during the normalized oscillation cycle are indicated.

## Discussion

### Time independent data

The study was performed to test the influence of using the device on trunk muscles, regarding their relationship to the biomechanically [19] and functionally defined muscle systems [20]. We expected continuous activation of the local MF with only little alteration from task characteristics, and strongly task related adaptive changes for the other global stabilizing and mobilizing trunk muscles. In this context, the results have to be considered as inconsistent with respect to the evoked activation characteristics. The role of RA, which belongs to the global mobilizer group, remained marginal for most situations. OE, which is considered to belong to the global stabilizer group also remained at insignificant amplitude levels which hardly cross 10 µV if considering the whole group.

Most mean MF SEMG amplitude levels were significantly higher in men than women (exceptions being fast frequency in vertical plane and left MF at slow and moderate frequency in horizontal plane). In contrast, women showed significantly higher OE amplitudes, except right sided OE for moderate and fast oscillation frequencies. Together with MF, here gender differences were obvious: OE amplitudes of the female subjects reached significantly higher levels in comparison with the investigated men. But the mean difference was rather small. SEMG amplitudes are subject to several technical [25-27] and physiological alterations. The distance between EMG source and detection site plays an important role: in general SEMG amplitudes are considered to be attenuated with increasing skin fold thickness [28]. Therefore, this slight shift might be interpreted as gender related in terms of different amplitude attenuation because the average BMI was significantly higher for the investigated men. However, if so this effect should be a general one, observable for all investigated trunk muscles. In contrast, ANOVA revealed higher amplitudes of MF muscle in men than women. Also for OI a slight but insignificantly higher level could be found for the male group. Assuming the comparably small number of investigated subjects, this observation likely could be verified with a higher number of cases (for details please see section "statistical considerations"). Therefore, for the found mean SEMG amplitude differences between both genders a general "skin fold related effect" can be excluded. Differences in fiber type composition can be excluded also, because of the same proportions in both genders [21]. But, fiber type proportion and functional cross sectional area of the respective fiber types are different issues. It is known that men have generally larger muscle fibers than women [29]. Furthermore, in contrast to men, women's Type I fibers are larger than their Type II fibers [21]. Anyhow, these differences are not able to evoke the observed differences, because both back and abdominal muscles are comparable in fiber type composition between genders. Differences on fiber type composition between front and back regions are known: abdominal muscles have higher proportions of Type II fibers in comparison to back muscles [30,31]. All these variations, should cause regionally related differences. But the observed distinctions between genders were related to certain muscles without regional association. Consequently, these differences have to considered as gender related with respect to differently organized coordination patterns of trunk muscles. Although, not mentioned as a result, obviously, side differences occurred for the abdominal muscles with right-sided predominance of RA and OE and left sided predominance of OI. Back muscle mean amplitudes were virtually symmetric. Since almost all subjects were right handed (one female ambidextrous, one male left-handed) these differences are most probably related to handedness. Data about trunk muscle amplitude levels in relation to handedness are not available from the literature, but other investigations also revealed asymmetrical amplitudes of shoulder muscles in relation to handedness [32].The occurring opposite predominance between both oblique abdominal muscles may be subject to measurement site: OI at the position it was actually measured shares fiber direction with the deep transverse abdominal (TA) muscle, which belongs to the local muscle group. Because cross talk cannot be eliminated completely [33], OI data are expected to partly reflect local muscle activation characteristics. During predominantly right sided hand use pronounced left sided compensation might be necessary. The known crossing of the thoracolumbar fascia [34] supports this hypothesis, but further investigations are necessary to highlight this specific issue.

**Time dependent data**

Oscillation plane did not affect abdominal muscle activation characteristics, but back muscle activation patterns differed considerably between both planes: In horizontal plane, mean amplitude appeared to be slightly higher for ES. In vertical plane, MF exhibited slightly higher mean amplitudes. In horizontal plane, both muscles were characterized by continuous activation, like all other investigated muscles. In vertical plane, this activation changed into a clear phasic pattern: Just after the top inversion point, during the initial downward movement of the pole ends an amplitude peak was exhibited. Therefore, the change of oscillation plane was able to evoke completely different activation characteristics which were found to be similar for both back muscles. Furthermore, relative time of the amplitude peak shifted forward with increasing frequency. Therefore, reaction times of MF and ES muscles also decreased, from 60 ms at 3 Hz via 45 ms at 4 Hz towards 13 ms at 4.5 Hz after the top inversion point. Most probably, this is a compensational mechanism for the electromechnical delay (EMD) which is known to be invariant in time for repetitive muscle activities [35,36]. The remaining time from back muscles amplitude peak till bottom inversion point stayed constant likewise: The resulting intervals were found to be 107 ms at 3 Hz, 97 ms at 3.5 Hz, and 98 ms at 4.5 Hz. These data are in accordance with data from the literature [37].

Since MF and ES muscles are assigned to either the local (MF) or the global (ES) systems [19], the similarities of MF and ES coordination patterns are surprising. As a local muscle, basically, a more continuous characteristic would have been expected for MF, but its peak amplitudes reached even higher amplitude levels when compared to ES. This also applies if looking at the amplitude normalized data, which showed a greater range for MF too (see figure 3). Even though one cannot expect to exactly measure activation characteristics of deep MF at the surface [38], a virtually identical pattern for ES during the downward moving phase of the pole ends could be observed [38]. During this phase most likely an eccentric contraction is evoked to compensate for the following downward directed force on the muscles at bottom inversion of the pole. Therefore, it is not the MF activation pattern, but the ES activation pattern, which is the most surprising result. Functionally spoken, the concordant pattern qualifies ES to act also with eccentric activation as usually assigned to global stabilizers [20]. In other words, as was stated already by other authors, there is no "key muscle" [39,40] but proper coordination of all involved trunk muscles is an important feature for adequate trunk stability.

The cyclic activation enforces feed forward activation mechanisms to compensate for the trunk flexing moment applied by the device. Delayed feed forward activation of local muscles [6] is accepted being a key factor in the pathogenesis of LBP. Their reaction time could be shortened as a direct result of repeated voluntary abdominal activation, independently from whether specific TA training or general abdominal activation was performed [41].

Therefore, the application of the device can be a contributing factor for the diagnosis, prevention and, possibly, the therapy of low back pain. The variation of oscillation planes and frequencies has considerable effect, most notably on back muscle coordination patterns.

The device comes in four different lengths, from 130 cm up to 190 cm in 20 cm steps, resulting in a frequency range from 2.5 Hz up to 7.5 Hz. This enables a wide range of applications from rehabilitation to fitness. Compared to other training devices the Propriomed® offers an reasonable alternative to train trunk muscle coordination.

## Limitations

### Physiological considerations

With the presented data we cannot evaluate if any strengthening effect of trunk muscles might be expected by using the device, because prior to the investigation no maximum voluntary contraction test (MVC) was performed. This was particularly chosen because of two reasons: First of all, it is extreme strenuous for the subjects to produce MVC data for all investigated trunk muscles separately. Furthermore, among others [42] motivational influences [43] have impact on the measured rms values. This becomes even more important if low back pain patients have to be investigated since their data will not reflect reliable MVC levels [44]. Secondly, investigations during gait, which is a common cyclic activity revealed reduced reliability if data were normalized according previously determined MVC levels [45]. Investigations during running, i.e. maximum cyclic effort, could prove peak amplitudes which exceed static MVC levels by far [46].

A recently published paper by Moreside and colleagues [24] investigating a similar but different device (bodyblade, natural frequency 4.5 Hz) revealed mean trunk muscle amplitudes that reached between 10% up to 60% of previously determined MVC levels of trunk muscles if oscillation amplitude was increased. One test situation was similar to vertical oscillations of our setup, but differed in detail. Therefore it is difficult to directly compare both results. In our investigation only the oscillation frequency was changed, with only little effect on amplitude levels.

The investigation was performed with healthy volunteers. Therefore, the question still remains unanswered if the observed patterns can be evoked also in patients with LBP. If the application of different oscillation frequencies is suited to improve spinal stability also needs separate investigations.

### Statistical considerations

In general, small sample sizes require larger effect sizes to be detected. This effect is independent from what kind of statistics is applied. Power analysis is not available for nonparametric statistical methods for dependent variables, therefore we approximated the necessary effect size of the data by calculating the power for dependent variables while using the paired t-test. Power analysis [47] revealed a required effect size (difference between mean values over variance of the differences) of 0.53 between groups, according the given sample size of 30 subjects for a common power level of 80%. With only 15 subjects per group, as was the case for testing gender related differences, the necessary effect size to be detected increases to 1.36. Therefore, considering the limited number of investigated subjects, the probability to interpret false positive results (second order statistical error) has to be regarded as very unlikely.

## Conclusion

In healthy subjects the device is able to induce predictable activation of all investigated trunk muscles. Oscillation frequency and oscillation plane had only little to negligible effect on mean trunk muscle amplitudes. Oscillation planes evoke different activation characteristics for the back muscles but not for the abdominal muscles. Abdominal muscles all showed continuous activation patterns. For the back muscles, during vertical oscillations, clear phasic pattern could be observed whereas in horizontal plane a continuous activation existed. The kind of cyclic alternating force application on the trunk manifested by holding an oscillating object might be used for diagnosis, prevention, and treatment of impaired back muscle function.

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**Suppliers**

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# 9 Originalarbeit 3

## Cyclic upper body perturbations caused by a flexible pole: Influence of oscillation frequency and direction on trunk muscle co-ordination

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**Abstract**

**Objective** The effect of a newly developed training device on trunk muscle co-ordination patterns was investigated.

**Design** A cross sectional survey of 30 healthy volunteers was executed. The task was to maintain position and motion of a flexible pole that was set into oscillation. Oscillations were applied at three frequencies (3, 3.5, 4.5 Hz), in horizontal and vertical directions. SEMG signals of five trunk muscles were measured. Co-ordination was assessed by calculating mean relative amplitudes and mean, as well as, grand averaged muscle ratios during oscillation cycle for front over back (F/B), internal over external oblique (OI/OE), and multifidus over erector spinae (MF/ES) muscles.

**Results** Vertical oscillation evoked increased mean MF relative amplitudes and mean F/B and MF/ES muscle ratios in comparison with horizontal oscillation. Grand averaged F/B ratio was phase shifted about 180° of relative cycle between both oscillation directions. With increasing oscillation frequency mean F/B ratio increased, but OI/OE ratio decreased. Amplitude range of grand averaged muscle ratios during oscillation cycle was augmented with increasing frequency.

**Conclusions** The gathered normative data can serve as an initial data basis for further investigations on low back pain patients. Relative to oscillation direction and frequency, demands on trunk muscles vary from primarily stabilizing towards mobilizing activities.

## 1. Introduction

For all physical activities the co-ordinated exertion of agonist and antagonist trunk muscles [7,30,36] is essential to maintain the necessary equilibrium between stability and mobility.

There are few motor task requirements in which only one of these modalities is activated almost exclusively. These are during static (i.e. isometric) situations when primarily stability has to be maintained [27] and during ballistic movements, which strictly require mobility. Such extreme modalities are rare; during everyday activities we do however count on the whole physiological range of demands between them. For this reason stability and mobility demands have to be well organized to prevent failure [15].

Technologies for quantifying what would be considered proper muscle co-ordination are still developing. So far, the best suited method to describe muscle activation processes of motor tasks in vivo is Surface Electromyography (SEMG). The method uses surface electrodes and is primarily applied for quantifying muscle activation in terms of magnitude, timing, and fatigue, to name the three main applications [8]. If the co-ordination of several muscles has to be evaluated the calculation of relative amplitudes or also muscle ratios is helpful [10]. With this type of data analysis, one of the main disadvantages of SEMG, the great inter-individual variability, is reduced. The resulting vector between the observed muscles can be used for the identification of muscle co-ordination. Anyhow, inter-individual differences of regional fat distribution cause remaining and unavoidable disparities between subjects.

Data exist that demonstrate corrupted coordination patterns in association with patient back pain. In addition to measured activation timing problems [3,22], the coordination pattern data demonstrated even more consistently, that the agonist-antagonist ratio was affected with increased antagonist activity in relation to agonist activation level [22]. This could be proven in a variety of static [13] as well as dynamic activation situations [33]. Considering these results, it seems the identification and characterization of corrupted co-ordination patterns will be a key factor in pathophysiological motivated diagnosis as well as therapy of disturbed spinal stability.

The most successful therapeutic intervention strategy so far, the spinal segmental stabilization program [31], aims at the normalization of the necessary co-ordination of the involved trunk muscles to ensure a proper stability- mobility equilibrium. For economic reasons the necessity of therapist patient interaction limits universal application. Therefore, it would be a great step forward to enable training of the respective trunk muscles with a high degree of predictability, and without the need for continual therapist instruction in a therapeutic setting.

If alternating forces are applied to the trunk in a changing yet harmonious pattern, hypothetically, all trunk muscles should be activated in kind. Furthermore, if no trunk movement is performed during this task aside from compensating these movements, stabilization as well as mobilization activity reflecting this harmony should be accomplished. This study was performed to evaluate the effect of such specific task attributes on trunk muscle co-ordination patterns while using a new device for trunk muscle training. This device consists simply of a flexible pole. This pole can be set into tunable frequencies of oscillation, resulting in quickly alternating forces on the trunk. In order to assess trunk muscle co-ordination patterns muscle ratios were calculated in healthy volunteers.

Since motor learning is a basic process to optimize movement performance, the whole investigation was carried out two times to assess the influence of possible habituation by using the device.

## 2. Methods

30 healthy volunteers participated in this study (15 women, age 23.1 ± 2.0 years, 15 men, age 25.5 ± 5.7 years). All subjects gave their written informed consent prior to the investigation. The procedures followed the ethical standards of the Helsinki declaration in its actual version. The device (Propriomed®, Haider Bioswing, Germany) consists of a flexible pole with adjustable weights (figure 1). It comes in four lengths and we chose the 170 cm length. Oscillation frequencies can be adjusted by changing the position of the weights. The applied positions of the weights resulted in frequencies of 3 Hz, 3.5 Hz, and 4.5 Hz. If set into oscillation and correctly handled there is only one oscillation node, located in the middle of the device. The manufacturer suggests an amplitude of swinging between 30 to 50 cm, which we adhered to. The device is equipped with a handle, located in the middle of the pole having a width of 20 cm that enables gripping with both hands simultaneously. After initial initiation of oscillations correct use of the device results in virtually no movement of the grip, except the small impulses that

are necessary to keep the pole ends moving. The task was to perform oscillations of the pole ends while holding the device in both hands, with flexed arms in front of the body. Oscillations were applied in horizontal and vertical directions. During horizontal oscillation the pole ends move forward and back, whereas in vertical oscillation the ends move up and down. After initiating a frequency, a stationary oscillation time of ten seconds was maintained for measurement. The six possible frequency and direction combinations were applied in random order, separately determined for every volunteer. Each single task was performed three times.

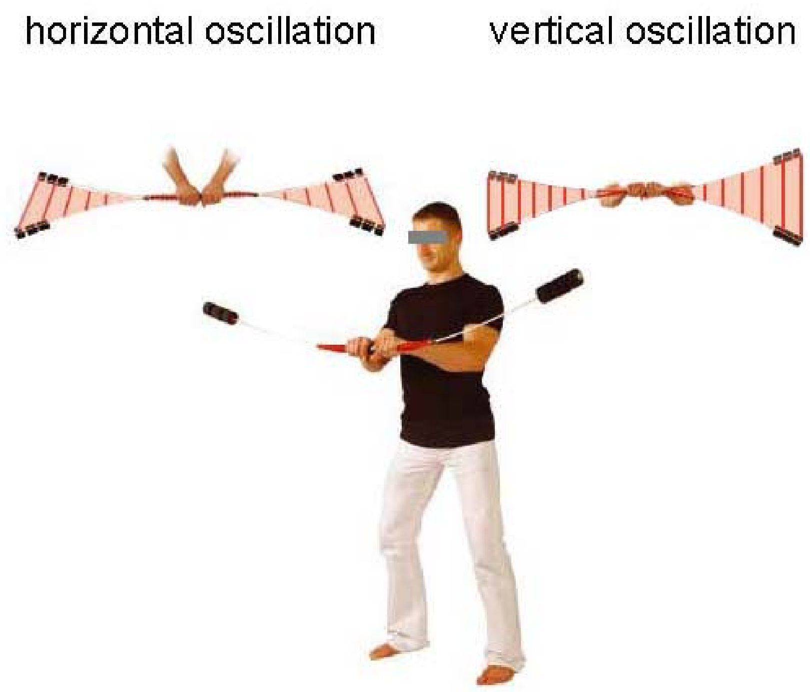


Figure 1 Schematic illustration of the operating mode of the device. Note, that it is a passive device, set into oscillation by the subject.

SEMG (5-500 Hz, gain 2500, Biovision, Germany, AD conversion: accuracy: 12 bit at 2000samples /s, DAQCard-AI-16E-4, National Instruments, USA) was measured from five different trunk muscles: straight abdominal muscle (RA), internal oblique abdominal muscle (OI), external oblique abdominal muscle (OE), multifidus muscle (MF), and erector spinae muscle (ES). According to international recommendations [17] muscles from both sides of the body were measured simultaneously (for detailed description of electrode location please refer to [2]).

The device was equipped with a bidirectional acceleration sensor to identify correct oscillation cycles with respect to timing and oscillation direction. For identification of correctly timed cycles the median value for all respective cycles of every single trial was calculated and only those cycles not exceeding 10% deviation from this value were used for further analysis. Since oscillation frequency was determined by the position of the adjustable weights this only occurred seldom, mostly caused by breaks due to unconscious change of oscillation direction. Correct oscillations with respect to oscillation direction were identified by comparing the two signal peaks of the accelerometer. Only cycles with a maximum time difference of 25 ms between these peaks were used for calculation. This criterion was established by analyzing optimally performed oscillations in horizontal and vertical directions, respectively. Cycles were time normalized to 100% relative cycle time with an accuracy of 0.5 % (201 points). The normalized cycle starts at rear inversion point for horizontal oscillations and top inversion point for vertical oscillations; 50% cycle time corresponds to frontal and bottom inversion point of the pole ends, respectively (see also figure 4).
SEMG data were band pass filtered between 20 and 300 Hz ($4^{th}$ order butterworth filter) and amplitude was expressed as root mean square (rms) using a moving window of 15 ms. Data from one task were averaged and the median value of the three repetitions was calculated as a grand average curve. For time independent amplitude parameters the mean value over the normalized cycle was calculated. From these data relative amplitudes were calculated, i.e. the ratio of each muscle according to the cumulative amplitude of all investigated muscles. According to the literature [10,35] trunk muscle co-ordination patterns were analyzed using the following ratios of SEMG amplitudes: front over back (RA+OI+OE/MF+ES: F/B), OI over OE (OI/OE), and MF over ES (MF/ES). These ratios were calculated for time dependent as well as for time independent data.
Because data were not normally distributed all statistical analyses were done using nonparametric tests: Wilcoxon's test for paired samples to test differences between oscillation directions and the Friedman ANOVA by ranks to identify influences of oscillation frequency. To test differences between the respective time dependent data, mean values of 5% cycle time were averaged using adequate tests as for the time independent data.

The whole investigation consisted of two single test sets with an intersession interval of one week for determination of intra- as well as intersession reliability of the data. To test reliability of the SEMG data mean amplitude levels were used and accuracy of task performance was assessed by the proportion of good cycles.

## 3. Results

### 3.1. Reliability analysis

Intrasession reliability for SEMG data was excellent for both investigation days, with intraclass correlation coefficient (ICC) levels ranging from 0.94 to 0.99 for day one and from 0.96 to 0.99 for day two. Task performance reached satisfactory levels: day 1 and day 2: 0.79. Intersession reliability (except RA for most tasks and left OI at 3 Hz, vertical direction: non satisfactory levels) reached ICC levels ranging from 0.53 to 0.92, i.e. satisfactory to excellent. In contrast, task performance showed training effects between the two investigation days, resulting in low reliability with an average value of 0.27. Mean proportion of good cycles reached levels of 78.6 % on day one and 89.2% on day two. Therefore, all data shown here will refer to the second investigation day.

### 3.2. Time independent data

Mean relative MF amplitudes of both sides differed between oscillation directions with levels in vertical direction exceeding those in horizontal direction (both MF: $p<0.001$). The influence of oscillation frequency remained small to negligible, but reached significance in some cases. During horizontal oscillation mean relative amplitudes significantly increased together with frequency for left RA ($p=0.025$) and right OE ($p=0.003$), but vice versa for right MF ($p=0.048$). For vertical oscillation, left MF ($p=0.025$) and left ES ($p=0.020$) relative amplitudes again significantly decreased with increasing frequency, whereas right OI ($p=0.048$) and right OE ($p=0.045$) showed significantly increasing mean relative amplitudes with increasing frequency (figure 2).

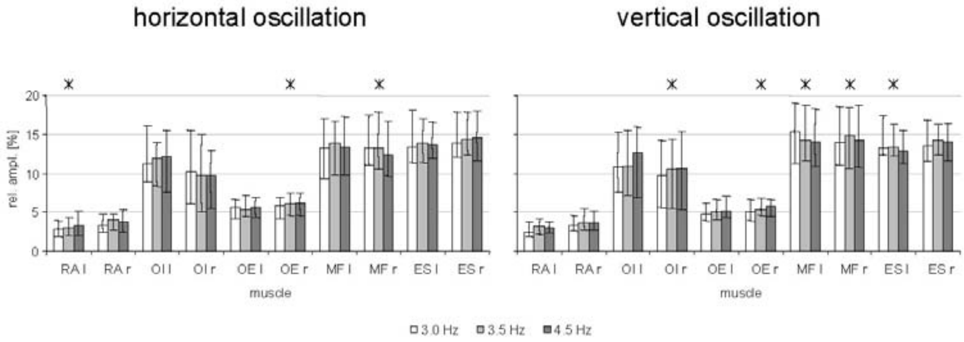


Figure 2 Mean relative amplitude of all investigated trunk muscles. *: significant differences between the three oscillation frequencies (p<0.05, Friedman ANOVA by ranks). Data are given as median values, and upper and lower quartiles, respectively.

F/B and MF/ES muscle ratios both showed similar differences depending on oscillation direction (figure 3): levels in vertical oscillation direction were significantly higher than in horizontal oscillation direction. OI/OE ratio remained unaffected by oscillation direction, but in horizontal direction increasing oscillation frequencies were accompanied by decreasing ratio levels (figure 4). In contrast F/B ratio increased with increasing oscillation frequency in vertical direction.

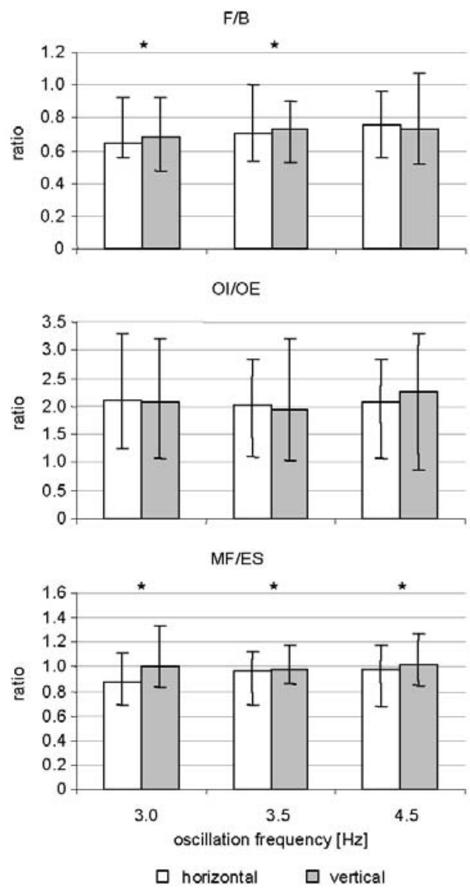


Figure 3 Influence of oscillation direction on mean muscle ratios over oscillation cycle. *: significant differences between oscillation directions ($p<0.05$, Wilcoxon Test for paired samples).

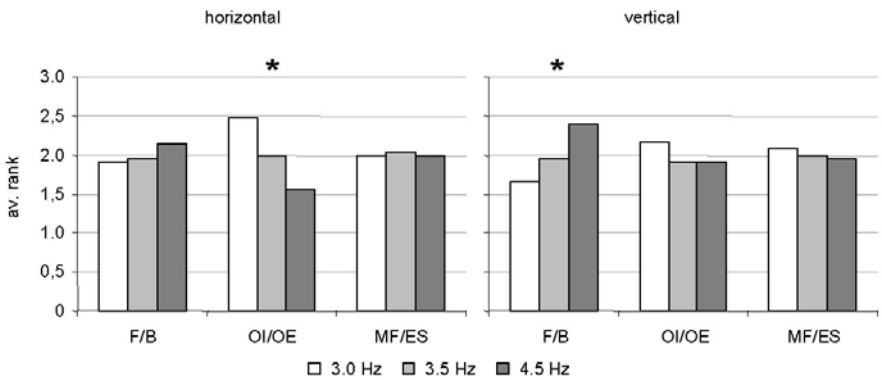


Figure 4 Results of the Friedman ANOVA by ranks: Influence of oscillation frequency on mean muscle ratios over oscillation cycle. *: significant differences between oscillation frequencies ($p<0.05$).

### 3.3. Time dependent data

Patterns of all calculated muscle ratios over the normalized cycle did not change considerably with frequency (figure 5). F/B ratio remained nearly balanced during horizontal oscillation, showing a slight change of predominant frontal activation around rear inversion point and back muscle predominance at frontal inversion point. During vertical oscillation this pattern became inverted and amplitude range was augmented: Back muscle predominance occurred around bottom inversion point whereas abdominal muscles prevailed around upper inversion point. Therefore, both patterns differed significantly. OI/OE ratio remained virtually unaffected by oscillation frequency as well as oscillation direction: During the forward or downward moving phase of the pole ends, muscle ratio levels reached highest values, whereas during the backward or upward moving phase, lowest OI/OE ratios were recorded. Even the lowest OI/OE levels revealed at least twofold rms levels for OI in comparison with OE. Highest OI/OE ratio levels reached about 3.5. MF/ES ratio again did not differ considerably in curve shape between either oscillation direction, but amplitude range was significantly larger during vertical oscillation direction. Around rear (upper) inversion point MF activation dominated, but around frontal (bottom) inversion point ES amplitude exceeded MF level. During horizontal oscillation, although the general pattern was similar, ES activation level exceeded MF level mostly during the normalized cycle time.

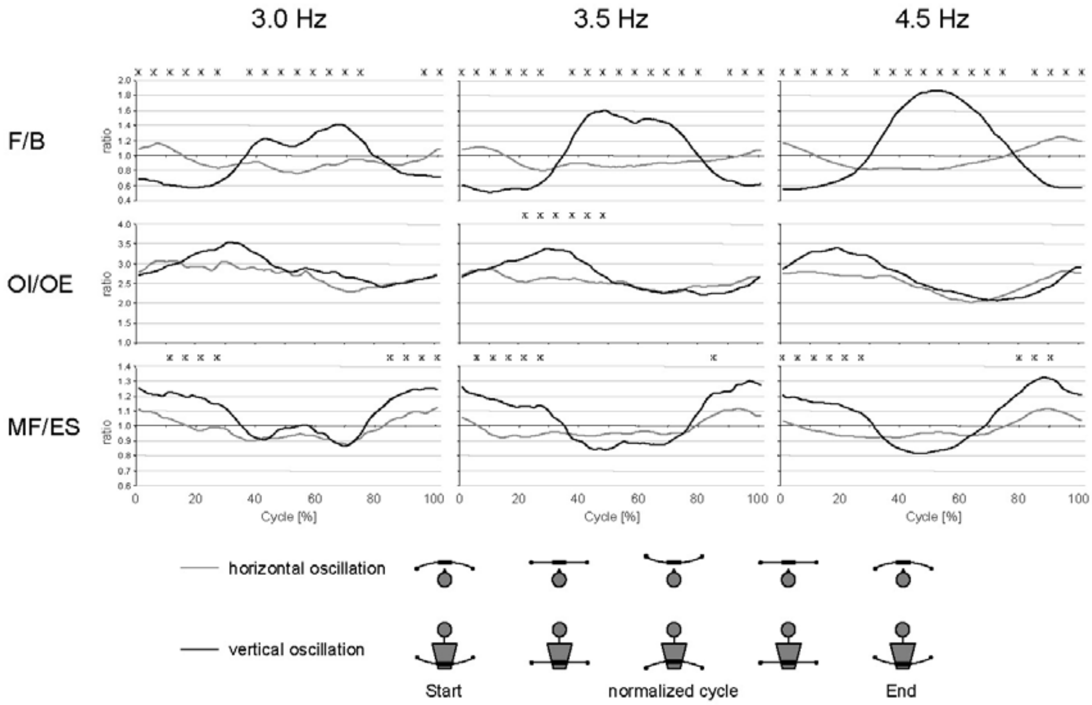


Figure 5 Grand averaged curves of calculated muscle ratios over oscillation cycle. Asterisks indicate significant differences between oscillation directions ($p<0.05$, Wilcoxon Test for paired samples). Positions of the pole ends are indicated with respect to normalized cycle phases.

## 4. Discussion

The device investigated was being tested with two goals: primarily to demonstrate if different frequencies and directions of oscillation clearly evoked changes in intra-individual muscle coordination patterns. The second purpose of the investigation was to determine how co-ordination patterns change over time with respect to use of the device during one session or between sessions.

In general one has to consider that muscle co-ordination patterns are subject to alteration because of motor learning [19], pain related influences [35], muscular fatigue [14], and actual force level as well as force direction [1].

Adaptation of intermuscular co-ordination towards optimization is a basic process of motor learning [4]. Therefore, the whole setup was performed two times. For this preliminary investigation reliability coefficients for mean SEMG data as well as the ratio of good cycles with respect to accuracy of timing and oscillation direction were computed. Surprisingly, SEMG data of these correctly performed cycles mostly were reliable but task performance itself was not. This indicates highly robust initiation of muscle function in terms of mean amplitude levels whereas task performance itself was subject to adaptation due to motor learning. In other words, if a person is able to use the device accurately, at least in healthy subjects, the observed mean amplitude ratios can be expected.

Muscular fatigue is also related to changes in muscle co-ordination patterns. These alterations are mainly due to a temporal loss of force capacity of the target muscles, which therefore have to be assisted by agonists [27]. This compensation process is induced by the attempt to bear the required task. Temporal functional failure of the main actors therefore has to be regarded as the underlying process of changed muscular co-ordination during fatiguing tasks. To prevent fatigue-related influences investigation tasks were randomized. This also avoided influences from an ordered, and thus predictable sequence of the applied frequencies. [29]. Therefore, the observed effects of changed co-ordination patterns with frequency and between oscillation directions were not subject to experimental procedure, but can be regarded as being originated by the respective task characteristics.

**4.1. Time independent data**

Although, mean relative amplitude levels of several muscles changed with oscillation frequency, the magnitude of these changes remained small to negligible. The general patterns were not altered substantially.

Interestingly, most changes of mean relative amplitude values occurred during vertical oscillation. Here, in general, increasing oscillation frequencies caused decreasing ratios of the back muscles whereas relative abdominal muscle amplitudes increased. Consequently, also F/B ratio increased with increasing oscillation frequency.

The only difference in mean relative amplitude level resulting from oscillation direction occurred for MF: Its mean relative amplitude during vertical oscillation reached significantly higher levels if compared with horizontal direction. Consequently, for the calculated muscle ratios differences due to oscillation direction occurred for F/B and MF/ES ratios. This points out to a general increase of stabilizing demands during vertical oscillation.

The influence of oscillation frequency on mean muscle ratio values revealed two contrary results: increasing F/B ratio during vertical oscillations was opposed by decreasing OI/OE ratio for horizontal oscillations with increasing frequency.

During horizontal oscillation mainly forward and backward directed forces act on the trunk. The higher the frequency the higher the strain level becomes in relation to more quickly alternating force patterns and higher inertial moments to reverse movement directions. OI and OE muscles are both classified as global stabilizing muscles [7]. However, OI at the position it was actually measured [25] may function as a local stabilizer. This region of OI shares fiber direction with transverse abdominal muscle (TA). Since muscles are already known to act functionally diverse in terms of activation patterns [20,24,32] a functional analogous behavior with TA activation towards a greater extent of stabilization function can be supposed for the measured portion of OI. Furthermore, cross talk from TA cannot be excluded [11]. Therefore, the slight but significant decrease of OI/OE ratio with increasing oscillation frequency during horizontal oscillation argues for an increase of active compensation strategy, caused by increased OE activation, rather than permanent stability behavior to maintain necessary core stability.

During vertical oscillation the significantly increasing mean F/B ratio with increasing oscillation frequency may also be caused by increased global muscle activation levels. The group of frontal muscles, is generally thought to consist of two global stabilizers (OI, OE) and one mobilizer (RA) and the two investigated back muscles are categorized as one mobilizer (ES) and one stabilizer (MF). According to Moseley [24], MF at surface level most likely behaves like a global stabilizer. With increasing frequency the increasing moments on the trunk require more global compensation.

Furthermore, abdominal muscles are known to be less powerful in comparison to their antagonists [9,16]. Their strain levels probably already reach higher levels during increasing oscillation frequencies. This is indicated by the fact that for all abdominal muscles a tendency towards increasing mean relative amplitudes could be observed with increasing oscillation frequency.

Therefore, a tunable effect of predominantly local muscle activation at low oscillation frequencies towards more active, global activation at higher oscillation frequencies can be assumed for the device. In this investigation only one of the four possible device lengths was tested. The used device length of 170 cm was chosen according to preliminary investigations which showed most consistent results for this particular device length. Oscillation frequencies for the different models range from 2.5 up to 7 Hz. For both, vertical and horizontal oscillations a resonance frequency of about 5 Hz has to be assumed for the human trunk [18,26,28]. The application of higher oscillation frequencies, therefore may change muscle co-ordination in a qualitative manner instead of the more quantitative changes which were observed during the actual investigation. Since we did not apply higher frequencies this remains hypothetical and should be investigated in future.

**4.2 Time dependent data**

Grand averaged ratios for F/B as well as MF/ES remained tonic during horizontal oscillation (F/B 0.8 to 1.2, MF/ES 0.9 to 1.1). This changed into a clear phasic behavior during vertical oscillations (F/B 0.6 to 1.8, MF/ES 0.8 to 1.3). OI/OE ratio was only slightly affected by oscillation direction (horizontal: 2.0 to 3.0, vertical: 2.0 to 3.5). Therefore, the change of force direction had a direct influence on superficial trunk muscle co-ordination. During horizontal oscillation the resulting force vector changed only gradually: the permanently downwards directed force, caused by the weight of the device was modulated by frontal (front inversion point) to backwards (rear inversion point) directed forces. During vertical oscillations the weight of the device was increased at bottom inversion point and reduced by the inertial forces, which are generated by the device at top inversion point. Therefore, during vertical oscillation only positive and negative vertical forces act on the trunk. Because not only force direction, but also force amplitude, differed between oscillation directions, it is unclear if the different force vectors per se are responsible for evoking a more phasic or tonic activation pattern of superficial trunk muscles.

For F/B dynamic ratio, a type of pattern reversal, i.e. a phase shift of about 180° of relative cycle time, could be observed if comparing the two oscillation directions. During horizontal oscillation the maximum frontal directed F/B vector occurred just before the rear inversion point. During vertical oscillation the frontal directed vector could be observed at bottom inversion point.

At first sight these results seem to be conflicting, as they suppose opposite effects on muscular co-ordination during comparable force vectors. Specifically, an expected high frontally directed force vector just before the highest backward directed moment during horizontal oscillation contrasts to a similarly directed force vector, unexpectedly occurring just at the highest downwards directed moment during vertical oscillation.

The analysis of quickly alternating EMG co-ordination patterns has to involve differentiating between the force resulting from the response to the quickly alternating forces on the trunk and the force produced by the muscles, i.e. the electromechanical delay (EMD). Taking into account the inertia moment of the trunk and the electromechanical delay (EMD), the most backwards directed force vector, indicated by the preceding peak of F/B ratio, occurs just at bottom inversion point [34] during vertical oscillation. The shorter time offset for the resulting forward directed F/B ratio in horizontal direction might be explained by the shorter EMD of trunk flexor muscles [34], but cannot explain it completely. As already stated, abdominal muscles are less powerful than back muscles [9,16]. Furthermore, increasing load on the trunk leads to increased stability demands [6]. Taking into account, that overall spinal stability has to be accomplished by highly co-ordinated activation of all trunk muscles, with no single muscle as the most important [5,23], strain level of abdominal muscles may reach higher levels if compared to back muscles. Their amplitude increase, therefore, exceeds back muscle amplitude increase, resulting in a changed F/B ratio.

Peak amplitudes for both oscillation directions were more pronounced with increasing frequency. The data also suggest the impression of an earlier occurrence of the amplitude peaks with increasing frequency if regarding the whole curve without considering the incision at peak ratio levels. Unfortunately, at slow and moderate oscillation frequencies peak ratio levels were split into double peaks, therefore this impression remains hypothetical. This amplitude split most probably may be due to changes of MF/ES co-ordination, since it is observable for this particular ratio as well but fails to be detected for OI/OE ratio. Just before bottom inversion point MF ratio relative to ES was shortly increased. This phenomenon disappears at higher frequencies. As already stated, the resonance frequency of the trunk occurs at 5 Hz [26]. The highest applied oscillation frequency therefore, may already evoke incipient qualitative changes of co-ordination patterns. This remains hypothetical since we did not apply higher oscillation frequencies.

The observable peaks for OI/OE as well as MF/ES ratios, although not affected substantially by oscillation direction, together with increasing frequency occurred at earlier relative cycle times. Since absolute cycle time gets shorter with increasing frequency, this effect can be considered as compensational for EMD. Incidentally, the EMD is known to remain constant during repetitive tasks [12,21]. With the observed shift, the absolute time distance between the occurring peaks and the corresponding cycle phases could in this way be kept constant [34]. Therefore, not only adaptation capabilities related to force direction, but also in terms of adequate timing, can be assessed by using the device. The gathered normative data can serve as an initial data basis for further investigations on low back pain patients. Furthermore, the robust and effective stimulation of active trunk muscle stabilization functions may help to prevent the development of back problems.

## 5. Conclusions

Quickly alternating forces on the trunk exhibit expectable trunk muscle activation patterns in healthy subjects. Time independent co-ordination patterns, i.e. muscle ratios are only slightly biased by oscillation frequency as well as oscillation direction. Temporal characteristics of muscle ratios are subject to change, mainly with respect to oscillation direction: F/B ratio is phase shifted about 180° of relative cycle time if horizontal oscillation is compared with vertical oscillation. Temporal OI/OE and MF/ES ratios are augmented during vertical oscillation, but are not substantially affected by oscillation direction. Occurring ratio peaks shift forward in relative cycle time with increasing oscillation frequency, most probably to maintain the necessary invariant time delay between electrical activation and force production. Furthermore, increasing oscillation frequencies seem to shift muscle exertion from stabilizing towards mobilizing muscles. The gathered data from healthy subjects may serve as a data basis for future clinical use. In future, use of the device may provide auxiliary therapy for improving trunk muscle co-ordination to enhance active trunk stabilization.

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## 10 Originalarbeit 4

### Evaluation of the EMG-force relationship of trunk muscles during whole body tilt

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**Abstract**

The study was aimed at the identification of the Electromyography (EMG)-force relationship of five different trunk muscles. EMG-force relationships differ depending on changes in firing rate and the concurrent recruitment of motor units, which are linear and S-shaped, respectively. Trunk muscles are viewed as belonging to either the local or global muscle systems. Based on such assumptions it would be expected that these functionally assigned muscles use different activation strategies.

31 healthy volunteers (16 women, 15 men) were investigated. Forces on the trunk were applied with the use of a device that gradually tilts the body to horizontal position. Rotation capability enabled investigation of forward and backward as well as right and left sideward tilt directions. Surface EMG (SEMG) of five trunk muscles was taken. Root mean square (rms) values were computed and relative amplitudes, according to the measured maximum amplitudes, were calculated individually.

Back muscles were characterized by a linear SEMG-force relationship during forward tilt. Abdominal muscles showed an S-shaped polynomial SEMG-force relationship for backward tilt direction. Sideward tilt directions evoked lesser SEMG levels with polynomial curve characteristics for all investigated muscles.

Therefore, the SEMG-force relationship possibly is also subject to force vector in relation to fiber direction.

## Introduction

To maintain the necessary equilibrium of stability and mobility of the spine trunk muscle co-ordination has to be adequately organized (Gardner-Morse and Stokes, 1998; McGill et al., 2003). During the development of this functional view, functional characteristics were assigned to all trunk muscles. Primarily this assignment was biomechanically determined (Bergmark, 1989) based on whether these muscles insert directly at the spine (local system) or transfer forces between thorax and pelvis (global system). Within this system local muscles are related to continuous, low level activation, independent from movement to ensure stability (i.e. deep multifidus muscle). Global muscles act phasically at medium to high intensities to produce movements (i.e. erector spinae muscle). More recently, this classification was expanded and the assignment of functional characteristics of global muscles became more differentiated: global stabilizers limit range of motion and therefore mainly act eccentric (i.e. oblique abdominal muscles). Global mobilizers perform concentric contractions to initiate movements (Comerford and Mottram, 2001).

Corrupted trunk muscle co-ordination was found to be related to low back pain (Panjabi, 2003). Therefore, trunk muscle function has been investigated intensively over the last two decades (Hides et al., 1994; Hodges and Richardson, 1996; Kankaanpää et al., 1996; Leinonen et al., 2003; Magnusson et al., 1996), mostly by using Surface Electromyography (SEMG) techniques.

Any activation of muscles is functionally driven, resulting in random activations of motor units (MU's, Basmajian and De Luca, 1985). Therefore, the resulting SEMG signal has a stochastic characteristic, also called "noise-like interference pattern" (McGill, 2004). It is correlated with global SEMG parameters like root mean square (rms) and median frequency (Bonato, 2001) to name the most common ones. These parameters are correlated with muscle function characteristics (Luttmann et al., 1996) and, therefore, provide insight into muscle function in vivo. Common SEMG applications are directed at the quantification of muscle co-ordination (Brown et al., 2007; Hodges and Richardson, 1999; van Dieen et al., 1996), identification of muscular fatigue (Luttmann et al., 1996) and the estimation of strain levels according to previously defined reference levels (Attebrant et al., 1995; Doorenbosch et al., 2005).

The latter application is used to determine if a certain load induces the predicted strain level, or, which often is of even more importance, to determine strain levels during load situations for which the load level cannot be determined. To approximate muscle strain levels reliable assumptions about SEMG amplitude – force relationships of the respective muscles are required. Early investigations already revealed differing SEMG amplitude – force relationships between muscles (Lawrence and De Luca, 1983). Further experiments could show a strong dependency of varying SEMG amplitude – force relationships from different recruitment strategies (Solomonow et al., 1990). A linear amplitude-force relationship was evident when all MU's were recruited up to 50% of maximum force and further increase of force resulted from increased firing frequency. If MU's together with increasing firing rates were concurrently recruited throughout the whole force range a nonlinear curve could be demonstrated (Solomonow et al., 1990).

A simulation approach (Fuglevand et al., 1993) including different maximum but fixed firing rates and recruitment ranges of all respective MU's revealed contrasting results: only if MU's were recruited over a broad force range, could a linear SEMG amplitude – force relationship be seen. If all MU's were recruited below 50% of maximum force nonlinear curve shapes were calculated, inconsistent with experimental findings. The authors could not output s-shaped curve characteristics.

Another known variable affecting the SEMG amplitude – force relationship is the muscle length – SEMG relationship (Anders et al., 2004) requiring identical muscle lengths during the normalization and the test situations to avoid false conclusions. Different positions of electrodes with respect to the endplate region also impact measured SEMG levels (Farina et al., 2002; Kleine et al., 2000).

According to the well-established assignment of trunk muscles to either the stabilizing or mobilizing muscle systems (Bergmark, 1989; Comerford and Mottram, 2001), we hypothesized each system would employ different strategies: Local muscles were expected to exhibit variation of firing frequency owing to their continuous activation characteristic (Comerford and Mottram, 2001) and, in contrast, mobilizing muscles were expected to additionally recruit more MU's to reach the high force levels necessary to limit as well as initiate movements (Comerford and Mottram, 2001).

According to what is known from the literature muscles of these two separate systems were expected to show distinct EMG-force relationships: A linear characteristic for local muscles and non-linear curves for the global muscles.

**Methods**

For this study 31 healthy volunteers (16 women, mean age 23.1 ± 4.9 SD, 15 men, mean age 25.7 ± 4.7 SD) were investigated. All subjects volunteered after signing written informed consent. The investigation was performed in a device for trunk muscle diagnosis and treatment (Centaur®, BfMC, Leipzig, Germany, figure 1).

Figure 1: Operation mode of the device for whole body tilt. In contrast to this schematic demonstration for standard applications, during the actual investigation arms were held crossed against chest.

This device applies forces on the trunk by tilting the whole body from neutral upright position. Subjects are fixed at their feet, thighs and hips, but the trunk remains unsupported. During the different tilt positions the subject simply has to stabilize his or her upper body in body axis. For this investigation subjects held their arms crossed against their chests. Exact body and arm positioning throughout the whole investigation was controlled by the examiner. During the complete investigation no active trunk movement needed to be performed; the gravitational forces were simply compensated. Force levels and directions were applied by tilting the device to angles of up to 90° (i.e. horizontal position) and rotational angles up to 360°, respectively (figure 1).

Gravitational forces are always vertically directed. Therefore, at 0° tilt angle no sideward torque at all is applied on the trunk. With increasing tilt angles torque levels increased up to 100% of the gravitational force at 90° tilt angle. This increase of torque levels follows a sine function. Depending on the actual rotation angle the resulting force vector on the trunk was backward, forward, and sideward directed, respectively.

The following positions were investigated: Forward and backward tilt at 5°, 10°, 20°, 30°, 45°, 60°, and 90° tilt angles. These angles correspond with normalized torque levels of 9%, 17%, 34%, 50%, 71%, 87%, and 100%. Although this graduation is not equally distributed, it reflects the whole range at adequately separated torque levels. Due to expected overload exposures left-sided and right-sided tilts (i.e. tilts in frontal plane) were applied only up to 71% of the whole tilt range (45° tilt angle, Denner, 1998).

To compare side tilts with forward and backward tilts the used range for sideward tilt directions was also applied for the sagittal plane. Therefore, for direct comparisons between the two tilt planes normalized torque levels are limited to 71% of the whole tilt range.

Positions were applied randomly, each consisting of three repetitions of 10 seconds duration to ensure stationary conditions for at least five seconds. SEMG (5 Hz – 700 Hz, Biovision, Wehrheim, Germany) of five trunk muscles was taken simultaneously from both sides of the body: Rectus abdominis muscle (RA), obliquus internus abdominis muscle (OI), obliquus externus abdominis muscle (OE), multifidus muscle (lumbar part, MF), and erector spinae muscle (longissimus, ES). Electrode positions were chosen according to international recommendations (Hermens et al., 1999) and are described in detail elsewhere (Anders et al., 2005). We used disposable Ag-AgCl electrodes (Arbo®, Neustadt/Donau, Germany) with a circular uptake area of 1 cm diameter and an inter-electrode distance of 2.5 cm. Data were stored on computer for further offline analysis (AD-conversion at 2000/s, DAQCard -AI-16E-4, National Instruments, Austin, TX). To avoid false results by quantifying possible offset values raw SEMG was centered by subtracting the mean value of every single channel and measurement from the respective data. Furthermore, data were band-pass filtered ($4^{th}$ order Butterworth filter, cut-off frequencies 20 Hz and 400 Hz, respectively). We also recorded the electrocardiographical activation (ECG) by using an additional electrode pair along heart axis. This channel was used to detect ventricular activation. Quantification of SEMG data was done for intermittent segments of a duration of 0.4 seconds whereas an interval of 0.1 seconds after the respective R-wave was maintained to avoid possible ventricular interference on SEMG data. By doing this about ten sections could be used for quantification for each single repetition. Root mean square (rms) was calculated for every single section and these values were averaged for each repetition. As a representative of the respective situation the median values of the three repetitions were used for calculations. Rms data from both sides of the body were pooled. Pooling in frontal plane was done using the respective rms data during contralateral tilt directions ( i.e. right sided muscles during left sided tilts pooled with left sided muscles during right sided tilts).

Since data were not always normally distributed amplitude levels at the respective tilt angles were compared using the non-parametric Wilcoxon test for dependent samples (SPSS®, Chicago, Il). All SEMG data will be displayed as median values and upper/lower quartiles, respectively.

The SEMG amplitude to force relationship was evaluated by using normalized data for both parameters. SEMG amplitudes are expressed as relative amplitudes according the occurring maximum SEMG level. Torque levels are displayed as relative torque levels according the applied torque range.

## Results

### forward/backward tilt

SEMG amplitude increase with increasing torque differed considerably between abdominal and back muscles. With increasing torque all abdominal muscles showed a nonlinear amplitude increase. In contrast the back muscles were characterized by a linear increase (figure 2). Since amplitude increase was similar for the abdominal muscles on the one hand and the back muscles on the other, SEMG data were pooled according muscle localization at the front and back side of the body, respectively (figure 3) to exemplify the observed characteristics.

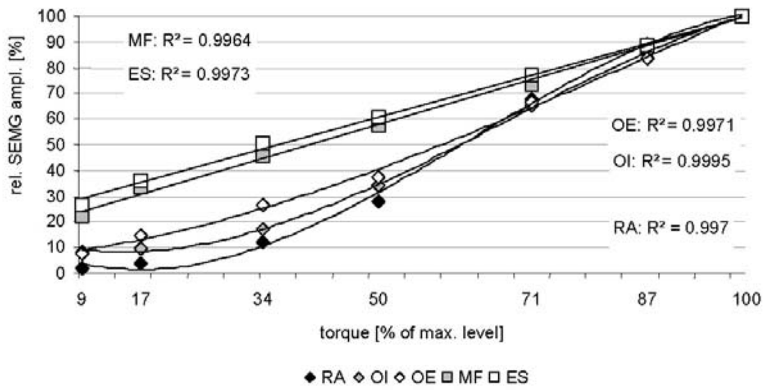


Figure 2: SEMG-force relationship of all investigated muscles. SEMG values at 100% of the applied torque range (i.e. 90° tilt angle) served as the reference level. Data are displayed as relative amplitudes. SEMG data for RA, OI, and OE were taken during backward tilt, SEMG data of MF and ES were obtained during forward tilt. Trend curves are related to the displayed median values, which are linear for the back muscle data and 3rd degree polynomials for the abdominal muscle data. $R^2$ values are given. Applied tilt angles are displayed as relative torque levels.

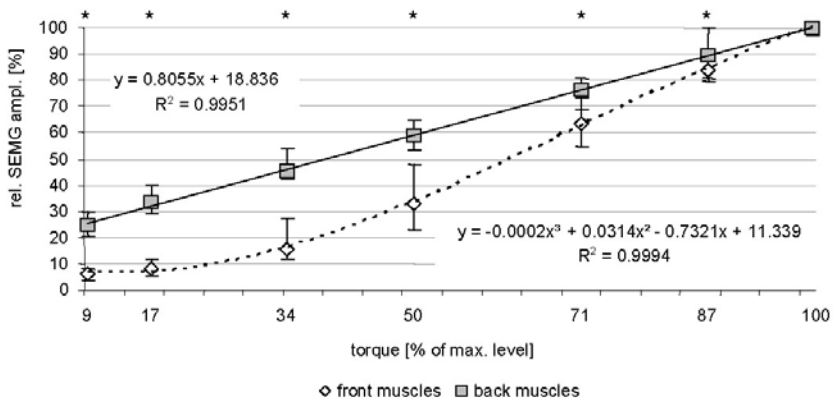


Figure 3: SEMG-force relationship (relative amplitude levels: median, upper/lower quartiles) of pooled abdominal (RA, OI, OE: abdominal muscles) and back (MF, ES: back muscles) muscle data. Trend curves are related to the displayed median values. The respective equations and $R^2$ levels are given. Applied tilt angles are displayed as relative torque levels. Asterisks indicate significant differences between the pooled abdominal and back muscles SEMG data of at least $p < 0.05$.

**sideward tilt**

Since, prior to the investigation strain level during sideward tilt directions was expected to excess maximum force capacity of the trunk muscles, tilt range was limited to 71% (i.e. 45 ° tilt angle) of the maximum applied toque range (Denner, 1998).

In comparison with forward and backward tilts, except OI, all trunk muscles showed significantly lower amplitudes during sideward tilt directions, respectively (figure 4). For OI a higher amplitude appeared for 9% and 17% of maximum torque for sideward tilt, but at 71% of maximum applied torque (45° tilt angle)

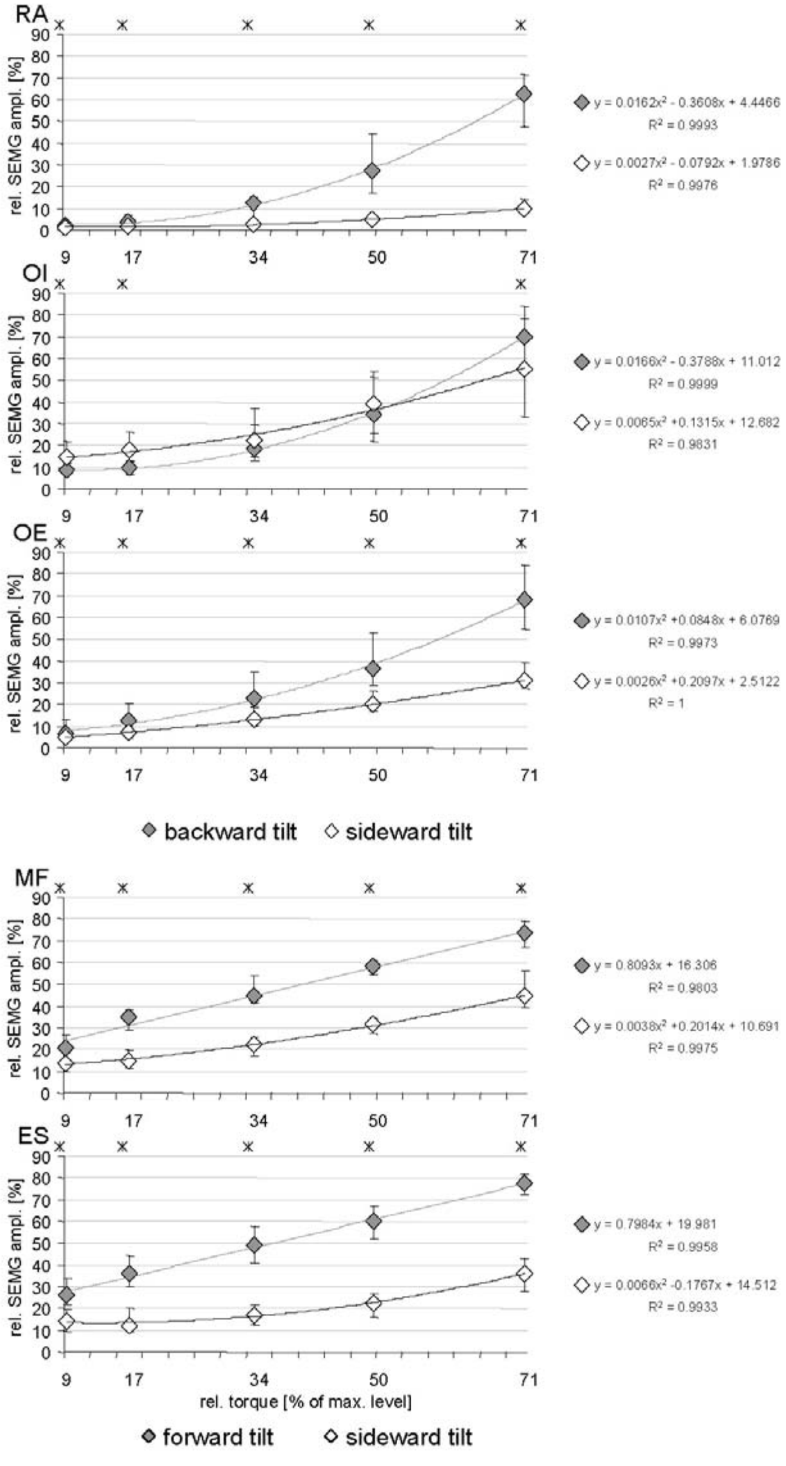


Figure 4: Trend curves for relative amplitudes (median, upper/lower quartiles) during backward and sideward tilt directions (abdominal muscles) and forward and sideward tilt directions (back muscles). For direct comparisons between planes relative torque levels are limited to 71% of maximum torque range (45° tilt angle). Trend curves are related to the displayed median values. $R^2$ values are indicated. Asterisks indicate significant differences of at least $p<0.05$.

backward tilt evoked higher amplitudes, just as for the other trunk muscles. For all abdominal muscles the amplitude differences increased with increasing torque levels. RA amplitude differences between tilt planes reached the highest levels:

backward tilt exceeded those during sideward tilt about six fold at 71% of maximum torque (45° tilt angle). For both back muscles amplitude differences between tilt planes remained virtually constant, at about two fold level in favor of forward tilt direction.

Additionally, the SEMG-force relationship differed between frontal and sagittal tilt planes: The observed differences for the abdominal muscles were mainly quantitative. Curve shape remained polynomial, but for tilts in frontal plane curve shape was changed towards a more linear characteristic, apparent in reduced multiplier values for $x^2$. Back muscles both changed to curve shaped from linear during forward tilt towards polynomial relationships during sideward tilt (figure 4).

## Discussion

All investigated muscles can be identified as either global mobilizing (RA, ES), global stabilizing (OI, OE, Comerford and Mottram, 2001), or local (MF, Bergmark, 1989). These muscle systems are considered to be different in function: Local muscles are permanently active, independent from movement, global stabilizers are characterized by eccentric activation in order to limit range of motion, and global mobilizers initiate movements, associated with concentric activation (Comerford and Mottram, 2001). In our investigation the muscles did not follow the expected functional patterns as was hypothesized. We expected a muscle-system related EMG-force relationship, but the found characteristics were related to abdominal and back arrangement of the investigated trunk muscles. In other words, the EMG-force relationship was not correlated with the assigned muscle function.

There already exist data by Lawrence and DeLuca about EMG-force relationships (Lawrence and De Luca, 1983). These findings demonstrated correlations for different muscles, which were also linear and non-linear, similar to our results. The findings were interpreted as being determined individually by the investigated muscles. Furthermore, they discussed different recruitment strategies, relative amounts of the different fiber types, cross-talk from neighboring muscles, fatigue, agonist-antagonist interactions, and viscoelastic properties. Because of the great inter-individual variability of fiber type composition the authors finally rejected any possible relationship between fiber type composition and EMG-force relationship. But in their data higher type II ratios coincided with a more nonlinear curve behavior of the respective muscles. Our data showed clear differences in curve shape between the investigated back and abdominal muscles. Although great inter-individual differences exist, fiber type composition seems to differ generally between these muscle groups: The proportion of type 1 fibers of the investigated abdominal muscles is estimated to be about 46%-58% (Haggmark and Thorstensson, 1979; Johnson et al., 1973), whereas back muscles contain about 54%-65% type 1 fibers (Johnson et al., 1973; Mannion, 1999). Thus, the proportion of type 2 fibers is explicitly larger in the abdominal muscles. Therefore, if muscle composition is considered, localization related pattern characteristics should be expected.

Solomonow and colleagues (Solomonow et al., 1990) demonstrated the influence of different recruitment strategies on EMG-force relationship. But also the range of force according to previously determined MVC levels influenced curve shape: The higher the range the more the curves developed a nonlinear shape.

As was expected from the literature (Keller and Roy, 2002) and also in our investigation, strain levels obviously differed between tilt directions. Backward tilt was much more strenuous for all subjects and, for some subjects, surely provoked maximum strain level at 90° tilt angle. Data from flexion - extension investigations are available showing relevant differences in force capability between abdominal and back muscles (Hakkinen et al., 2003). These data cannot be compared directly with our data, because of the known force – length relationship of muscles (Maganaris, 2001; Mannion and Dolan, 1996; Rassier et al., 1999). Furthermore, in general experimental data cannot directly be transferred between different setups (Hupli et al., 1997). To avoid influences from force production capacity and to directly compare positions in sagittal and frontal planes data of all abdominal muscles were plotted again, but range was limited to 45° tilt angle (71% of possible maximum torque level, figure 4). Since all tasks were randomly applied, systematic habituation errors can be excluded for the data.

Although, equal tilt angles in frontal plane provoked much less strain for all muscles, curve shape remained nonlinear for the abdominal muscles. But a tendency towards a lesser extent of nonlinear components was apparent. Unexpectedly, curves changed shape towards nonlinear characteristics for the investigated back muscles at sideward tilt. Therefore, the EMG-force relationship of a certain muscle seems to be influenced not only by force range, but also by force vector. In other words, the identification of a certain EMG-force relationship seems to be subject to change dependent on which angle an acting force acts relative to fiber direction of a certain muscle. This hypothesis is supported by the fact that in both oblique abdominal muscles sideward tilt directions resulted in a reduced non-linearity of the EMG-force relationship. This observation is indicated by reduced multipliers for $x^2$ of the respective equations (see figure 4). However, compared with sagittal tilt directions during tilt in frontal plane all muscles were activated at lower strain levels, indicated by lesser SEMG amplitudes. Therefore, also influences of the applied force range are possible reasons to explain this phenomenon (Solomonow et al., 1990).

Since we never reached maximum contraction levels for the back muscles the question remains unanswered if force range of isometric voluntary contractions may impact their curve shape.

Proper measurement of SEMG amplitudes requires attention to detail and is subject to numerous influences. To avoid fatigue related errors positions were randomized and sufficient breaks were kept between repetitions and between different positions. SEMG amplitudes of individuals are mainly influenced by individual strain level and detection distance, a direct influence of the thickness of the subcutaneous fat layer. This problem especially arises if subjects change posture throughout an investigation possibly causing varying degrees of skin attenuation (Mannion and Dolan, 1996). Since in the actual investigation body posture remained constant, errors from changed source-electrode distances can be excluded. Also muscle length remained unchanged, which is known to affect force production (Mannion and Dolan, 1996) and SEMG amplitude levels as well (Anders et al., 2004).

**Conclusions**

The recruitment strategies of abdominal and back muscles are different while counteracting backward and forward tilts. An ideal linear EMG-force relationship could be observed for back muscles, if loaded with extension forces. This correlation changed into a non-linear relationship if sideward flexion forces were applied. Abdominal muscles were characterized by a non-linear EMG-force relationship during application of flexion forces. For sideward flexion forces this characteristic became less nonlinear.

The assignment of muscles to either the local or the global muscle system does not correlate with these findings. A correlation between fiber composition, strain level and force vector relative to fiber direction and curve shape, or rather the used recruitment strategy, is likely, but cannot be definitively proven using the data gathered. In other words, the results argue for a flexibility of competing attributes: Local versus global, or functionally attributed versus task-related requirements.

**Conflict of interest statement**

Hereby all authors declare that neither personal nor economic relationships exists with the manufacturer of the device and that they did not receive any sponsoring that could have influenced their work.

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# 11 Originalarbeit 5

## Gender specific activation patterns of trunk muscles during whole body tilt

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**Abstract**

Gender specific differences as evidenced in both anthropometric data and physical performance of healthy persons have been broadly demonstrated. Recently advancements in surface electromyography (SEMG) have shown possible differences in men's and women's muscle coordination patterns. However, quantitative information about gender related muscle co-ordination patterns are rare. This investigation was carried out to both verify if trunk muscle SEMG amplitude-force relationship differs between men and women and refine techniques of measurement and data analysis using SEMG. 31 healthy volunteers (16 women, 15 men) were investigated during whole body tilt at angles from 5° to 90° (from quasi vertical to horizontal position). Subjects had to maintain body in body axis while their lower body was fixed and the upper body remained unsupported. SEMG was taken from five different trunk muscles of both sides simultaneously. At corresponding tilt angles women exhibited higher amplitude levels of their abdominal muscles in comparison to men, who were characterized by higher back muscle amplitudes. Abdominal muscles showed a non-linear SEMG amplitude-force relationship but differed between genders with more linearity in women. Back muscles showed a linear amplitude-force relationship with no differences between genders. Women were characterized by higher levels of co-contraction of all investigated muscles. The data are in accordance with histological investigations, which already proved specific fiber distribution patterns in both abdominal and back muscles and gender related differences in relative area of Type 1 fibers of back muscles. The observed differences in SEMG-force

relationship for the abdominal muscles remain hypothetical because of lack of histological information

## Introduction

Human beings belong to the group of animals displaying more pronounced morphological differences between the genders. Throughout the animal kingdom sexual dimorphism is strongly related to differing reproductive roles. Morphological investigations have demonstrated that movement characteristics typically differ most in connection with reproductive differences. Focusing on biomechanics, especially in mammals, hip architecture differs between genders. This distinction is not locally limited, but has consequences on trunk (Marras et al. 2001) and leg biomechanics (Kerrigan et al. 1998). In humans, gait patterns have been shown to differ between women and men (Smith et al. 2002).

In addition to movement differences relating to structural differences human bone density (Cooper et al. 1992) and muscle mass differ between genders (Janssen et al. 2000). The latter results from the larger cross sectional area (CSA), mainly of Type 2 fibers in men (Miller et al. 1993; Simoneau and Bouchard 1989). These gender differences were found to be more pronounced for the upper body compared to lower body regions (Gallagher and Heymsfield 1998; Janssen et al. 2000) resulting in unequal proportions of upper body mass (Denner 1998; Zaciorskij et al. 1984) between women and men. Fiber type distribution, on the other hand, seems to be similar between genders (Staron et al. 2000; Toft et al. 2003). Therefore, investigating human muscle function, the found gender differences in maximum force (Miller et al. 1993; Mital and Kumar 1998) and fatigability (Clark et al. 2003b; Mannion et al. 1997) rather are caused by quantitative (i.e. CSA) than qualitative (i.e. fiber type distribution) conditions.

In an earlier surface EMG (SEMG) investigation, we found different shoulder muscle co-ordination patterns in women as compared to men (Anders et al. 2004). Since shoulder anatomy does not differ between genders (Bonsell et al. 2000) this result argues for different strategies: Women tended to use stabilizing and also antagonist muscles more in comparison to men. Recently, however, gender differences could be proven for hip and thigh muscles during single leg landing (Zazulak et al. 2005), but, the results were contrariwise: Women tended to use their stabilizing hip muscles less, and their mobilizing rectus femoris more, as compared to men. Therefore, no general strategy of more intense co-contraction can be assigned to one gender up to now. Nowadays trunk muscle co-ordination is proposed to be the key factor in the etiology and pathogenesis of low back pain

(Hodges and Richardson 1996). For this reason different aspects of co-ordination have already been investigated (Arendt Nielsen et al. 1996; Hodges and Richardson 1998; Panjabi 1992). Until now, however, no study has posed the question of the existence of gender-specific trunk muscle activation patterns in terms of their amplitude-force relationship.

Physical training can induce changes in fiber type size, leading to a decreased EMG/force ratio during voluntary contractions (Hakkinen and Komi 1983). Since relative CSA of the respective fiber types differ between genders (Miller et al. 1993; Simoneau and Bouchard 1989) different amplitude force relationships can be expected.

The investigation was performed in a newly developed device for trunk muscle diagnosis and training. Therefore, the question had to be answered, if gender specific normative values should be used for later applications.

Thus, the aim of this investigation was to verify if gender specific trunk muscle activation characteristics are detectable during the execution of well-defined tasks. Such a well-defined task was provided by a device for trunk muscle diagnosis and training. In this device forces were applied by tilting the whole body. No intentional movement was required by the subjects except retention of upright posture.

**Methods**

31 healthy volunteers (16 women, mean age 23.1 ± 4.8 SD, 15 men, mean age 25.7 ± 4.7 SD) took part in this investigation (for detailed group characteristics see table 1). All subjects volunteered after signing written informed consent. Volunteers were recruited from a university setting. They were all investigated by the same physician to exclude possible limitations in physical capacity or orthopedic abnormalities. All subjects were free of back or any other pain. The investigation was performed with the aid of a newly developed device for trunk muscle diagnosis and treatment (Centaur®, BfMC, Leipzig, Germany, figure 1). Using this device, metered forces on the trunk can be applied by tilting the whole body from neutral upright to horizontal position or any position in between. Subjects are fixed at feet, thighs, and hips, while the trunk remains unsupported. During tilt, the person stabilizes his or her upper body in body axis. The device is also able to rotate the fixed subject up to 360°, therefore any tilt direction can be realized. No active movements are necessary because subjects are simply

Table 1: Group characteristics. Data are displayed as mean ± SD. Statistics were calculated using student's t-test for independent samples. Relative differences between mean values are given.

| | female (n=16) | male (n= 15) | p | relat. Diff. |
|---|---|---|---|---|
| age (y) | 23.1 ± 4.8 | 25.7 ± 4.7 | n.s. | 10.4 % |
| height (cm) | 170.3 ± 4.6 | 183.0 ± 5.8 | <0.001 | 6.9 % |
| weight (kg) | 57.0 ± 4.7 | 77.8 ± 8.8 | <0.001 | 26.7 % |
| BMI(kg/m²) | 19.7 ± 1.4 | 23.2 ± 1.6 | <0.001 | 15.2 % |
| UBM (kg) | 39.1 ± 2.3 | 51.6 ± 5.3 | <0.001 | 24.3 % |
| UBT (Nm) | 160.2 ± 13.4 | 226.7 ± 29.6 | <0.001 | 29.3 % |

UBM: upper body mass (including trunk, arms and head), calculation according Denner and Zaciorskij (Denner 1998; Zaciorskij et al. 1984)
UBT: upper body torque at 90° tilt angle (horizontal position), body proportions taken from Greil (Greil 2001), UBM was considered as a point mass at 50% of upper body height

figure 1: Operating mode of the device. Please note that during the actual investigation arms were held crossed against chest.

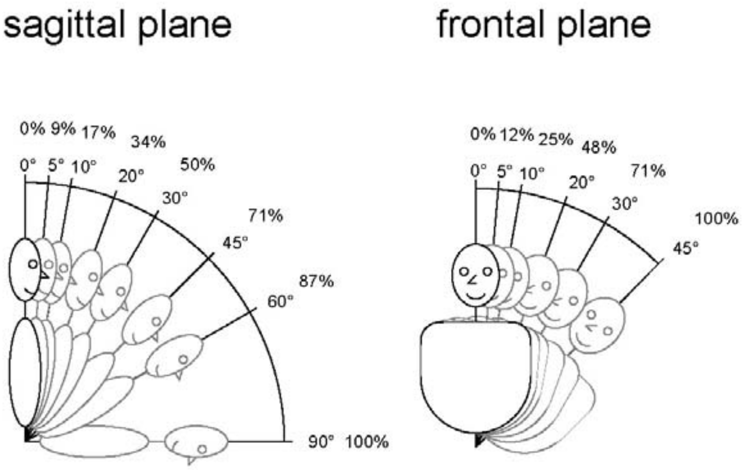


figure 2: Schematic illustration of the applied tilt angles and the respective relative torque levels for forward tilt in sagittal plane and left sided tilt in frontal plane.

compensating for gravity. To ensure consistent conditions throughout the investigation, subjects held their arms crossed against their chests. This also averted any co-contraction initiated by arm movements. Exact positioning throughout the whole investigation was controlled by the examiner. The following positions were investigated (see also figure 2): forward and backward tilt at 5°, 10°, 20°, 30°, 45°, 60°, and 90° tilt angles (sagittal plane); left-sided and right-sided tilt at 5°, 10°, 20°, 30°, and 45° tilt angles (frontal plane). Tilt angles can be converted into relative torque levels by applying a sine function. Therefore, within the range of applied forces in sagittal plane, these angles correspond with relative torque levels of 9%, 17%, 34%, 50%, 71%, 87%, and 100% of the applied torques. Although these graduations are not equally distributed they reflect the whole range at adequately equal separated torque levels. In frontal plane the respective relative torque levels correspond with 12%, 25%, 48%, 71%, and 100% of the applied tilt range. Due to the fact that in frontal plane tilt range was limited to 45°, relative torque levels cannot be compared between planes. The different tilt positions (i.e. tilt direction and angle) were applied randomly, each consisting of three repetitions of 10 seconds duration to ensure stationary conditions for at least five seconds. To prevent muscular fatigue at least 30 second of rest were maintained between the three subsequent repetitions. Main force directions were defined according to the main muscle functions - backward and contralateral sideward tilt directions for abdominal muscles and forward and contralateral sideward tilt directions for back muscles. For the respective sideward tilt directions, corresponding tilt angles for all contralateral tilt directions were used (i.e. left internal oblique muscle (OI) for right sided tilt and right OI for left sided tilt). The same principle applies for opposite force directions, but vice versa. Therefore, for opposite force directions back muscles were analyzed during back tilt, abdominal muscles during forward tilt and all muscles during the respective ipsilateral sideward tilt directions (i.e. left OI for left sided tilt and right OI for right sided tilt).

SEMG (5 Hz – 700 Hz, Biovision, Wehrheim, Germany) measurements of five trunk muscles were taken simultaneously from both sides: rectus abdominis muscle (RA), obliquus internus abdominis muscle (OI), obliquus externus abdominis muscle (OE), multifidus muscle (lumbar part, MF), and erector spinae muscle (longissimus, ES). These muscles were chosen because they reflect trunk stabilization characteristics well (McGill et al. 2003; van Dieen et al. 2003). Due to the use of SEMG transverse abdominal muscle was not measurable. Furthermore, since arms were held crossed against the chest additional influences from latissimus dorsi muscle could be excluded .

Electrode positions were chosen according to the literature and the respective international recommendations (Hermens et al. 1999; Ng et al. 1998). They are described in detail elsewhere (Anders et al. 2005).

Disposable Ag-AgCl electrodes (Arbo®, Neustadt/Donau, Germany) with a circular uptake area of 1 cm diameter and an inter-electrode distance of 2.5 cm were used. Data were stored on computer for further offline analysis (AD-conversion at 2000/s, DAQCard -AI-16E-4, National Instruments, Austin, TX, accuracy: 1µV/bit). Raw SEMG was centered and band-pass filtered ($4^{th}$ order Butterworth filter, cut-off frequencies 20 Hz and 400 Hz, respectively). Since ECG was also measured, ventricular activation could be detected and, therefore, eliminated.

Root mean square (RMS) was calculated at intervals of 400 ms, starting from 100 ms after the detected R wave to prevent influences from ventricular activity. Depending on heart rate and effective measurement time 5 to 20 intervals were averaged for every repetition. All repetitions were pooled by calculating median values. Data from both sides were pooled as well. RMS amplitude levels were normalized separately according to the maximum level in sagittal and frontal planes, which virtually always occurred at 90° in sagittal plane and 45° in frontal plane, respectively. Therefore, during forward and backward tilt directions 100% amplitude corresponds with 90° tilt angle (100% relative torque) and during sideward tilt directions 100% amplitude corresponds with 45° tilt angle (100%

relative torque). Amplitude levels were compared using the non-parametric Mann-Whitney test for independent samples (SPSS®, Chicago, Il). All SEMG data are displayed as median values and upper/lower quartiles, respectively.

## Results

### Main force direction:

#### Forward/backward tilt directions

Between genders almost all RMS amplitude levels for the abdominal and MF muscles differed (table 2): During backward tilt women showed higher RMS amplitudes for all investigated abdominal muscles, whereas men were characterized by higher RMS amplitude levels of MF during forward tilt. ES RMS amplitude levels were characterized by a tendency towards higher levels in women, significant at 45° and 60° tilt angles. In terms of relative amplitude levels women showed higher levels for abdominal muscles in most positions (RA: 5°-30°, OI and OE: 5°-45° tilt angles, figure 3). For the investigated back muscles no systematic differences in relative amplitude levels could be detected.

For subjects as a whole, amplitude-force curve shape was different between abdominal and back muscles, but specific differences existed between genders: Abdominal muscles showed a non-linear shape whereas the investigated back muscles offered a virtually perfect linear amplitude force relationship. The non-linearity of the abdominal amplitude-force relationship was more pronounced in men compared with the women.

#### Sideward tilt directions

In contrast to tilt in the sagittal plane, during tilt in frontal plane detectable differences could not be found for all investigated muscles (table 2). As in backward tilt, RA was characterized by slight but significantly higher RMS amplitudes in the women's group, but relative amplitudes did not differ systematically. For OE, again as well, women always showed higher RMS amplitudes. Its relative amplitudes only differed at 5° and 10° tilt angles. OI did not show any gender differences. For the back muscles gender differences appeared only for ES muscle: at 5° and 10° tilt angles. ES showed significantly higher RMS amplitudes in the men's group, but relative amplitudes did not differ. MF RMS amplitudes during sideward tilt did not differ at all.

Curve shape of amplitude-force relationship was non-linear for all investigated muscles with no essential differences between genders.

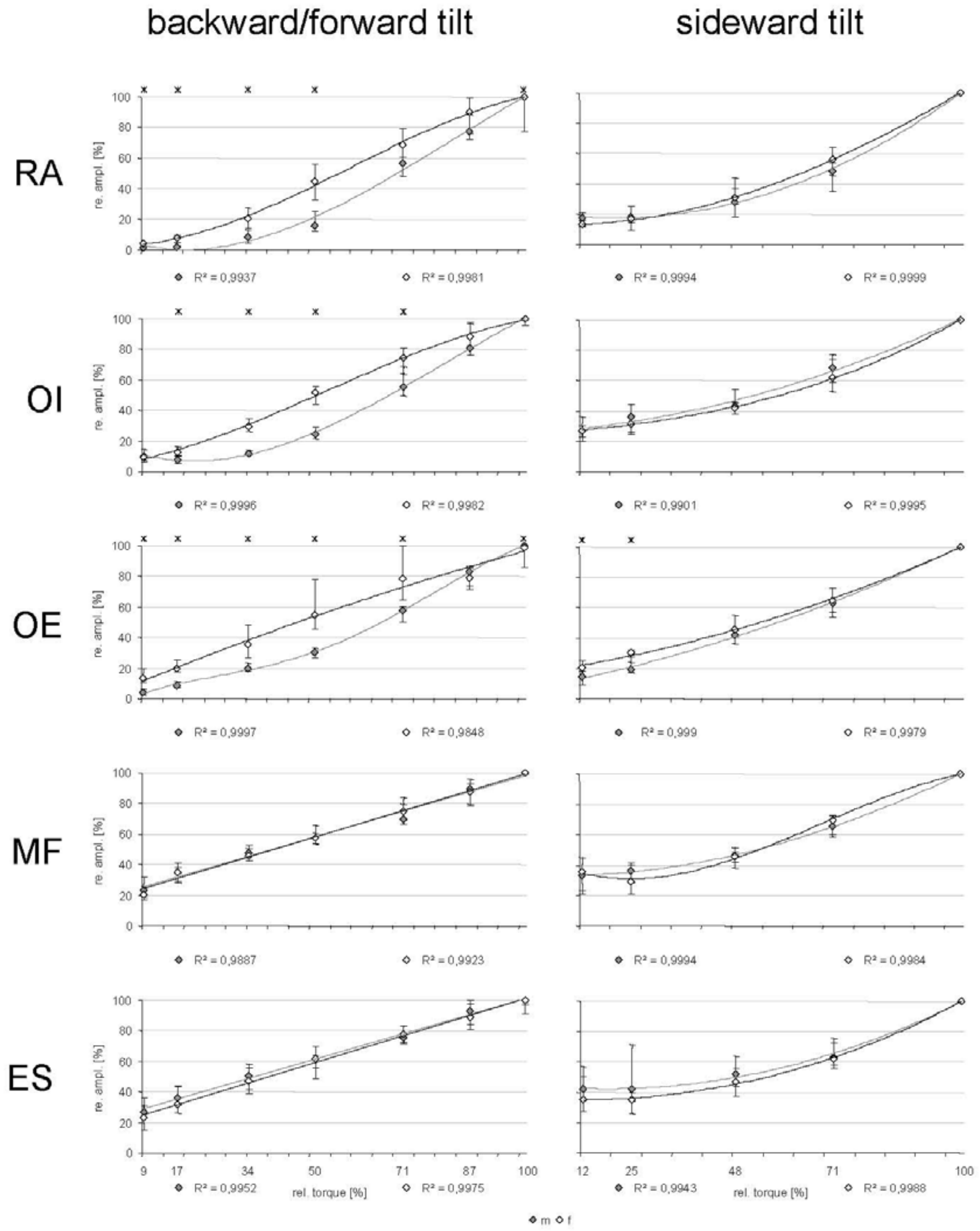


figure 3: Relative torque levels of trunk muscles in main force directions (Median, upper/lower quartile range). Note that relative torque levels differ between planes: In sagittal plane 100% corresponds with 90° tilt angle, in frontal plane 100% corresponds with 45° tilt angle. Asterisks indicate gender differences of at least $p<0.05$. $R^2$ values of curve fits are given.

Table 2: SEMG amplitudes (Median, (upper/lower quartile range)) during whole body tilt in main force directions.

| tilt angle (rel. torque) | | RA | | p | OI | | p | OE | | p | MF | | p | ES | | p |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| forward/backward tilt | | | | | | | | | | | | | | | | |
| 5° (9%) | m | 2.2 | (0.7/0.5) | 0.000 | 10.2 | (12.7/2.2) | n.s. | 4.1 | (4.3/0.8) | 0.000 | 25.0 | (10.5/6.4) | 0.009 | 31.4 | (4.6/3.3) | 0.006 |
| | f | 7.5 | (6.6/2.5) | | 15.8 | (4.6/2.2) | | 18.6 | (9.7/6.2) | | 17.8 | (3.3/2.0) | | 19.4 | (5.3/2.8) | |
| 10° (17%) | m | 2.4 | (1.9/1.1) | 0.000 | 9.9 | (1.9/3.2) | 0.000 | 13.0 | (2.2/4.4) | 0.000 | 42.7 | (5.1/8.4) | 0.003 | 43.5 | (4.0/10.8) | 0.048 |
| | f | 18.2 | (12.5/9.3) | | 22.2 | (12.0/3.2) | | 27.9 | (11.4/4.6) | | 28.7 | (4.9/4.0) | | 30.6 | (5.0/5.7) | |
| 20° (34%) | m | 10.7 | (15.3/4.4) | 0.001 | 18.1 | (5.2/4.4) | 0.000 | 27.1 | (7.3/4.9) | 0.000 | 55.6 | (14.5/9.2) | 0.005 | 52.7 | (19.0/12.1) | n.s. |
| | f | 48.1 | (31.5/14.6) | | 50.4 | (18.5/4.4) | | 49.4 | (25.0/10.7) | | 41.3 | (8.5/3.3) | | 39.7 | (15.4/4.9) | |
| 30° (50%) | m | 32.3 | (9.3/17.4) | 0.000 | 38.0 | (3.6/3.8) | 0.000 | 38.4 | (15.7/7.4) | 0.000 | 63.5 | (19.8/4.5) | 0.002 | 67.9 | (18.9/21.7) | n.s. |
| | f | 101.9 | (58.3/39.8) | | 93.3 | (32.0/3.8) | | 76.2 | (31.8/18.8) | | 55.7 | (4.8/8.0) | | 54.2 | (13.3/12.0) | |
| 45° (71%) | m | 135.0 | (9.6/36.0) | 0.020 | 83.0 | (34.0/9.9) | 0.014 | 81.7 | (19.4/21.7) | 0.016 | 89.6 | (22.3/13.8) | 0.014 | 82.1 | (27.1/26.1) | n.s. |
| | f | 172.0 | (67.0/52.0) | | 136.2 | (31.6/9.9) | | 118.2 | (24.7/34.5) | | 66.8 | (6.6/5.8) | | 69.9 | (11.1/16.3) | |
| 60° (87%) | m | 145.2 | (25.3/47.6) | 0.030 | 126.6 | (20.1/19.2) | 0.037 | 109.1 | (26.7/22.9) | n.s. | 107.1 | (15.2/9.5) | 0.004 | 91.9 | (42.4/21.8) | n.s. |
| | f | 207.9 | (88.4/67.3) | | 162.6 | (39.6/19.2) | | 110.4 | (23.5/18.8) | | 79.1 | (11.3/5.7) | | 80.5 | (19.6/19.2) | |
| 90° (100%) | m | 188.6 | (55.3/52.6) | n.s. | 156.9 | (26.7/20.1) | n.s. | 136.3 | (34.7/26.3) | n.s. | 113.5 | (15.7/10.6) | 0.004 | 107.9 | (36.6/37.4) | n.s. |
| | f | 238.9 | (77.3/69.8) | | 181.7 | (55.5/20.1) | | 135.0 | (15.1/37.3) | | 93.7 | (9.3/14.7) | | 91.5 | (13.1/20.1) | |
| sideward tilt | | | | | | | | | | | | | | | | |
| 5° (12%) | m | 2.2 | (0.6/0.4) | 0.002 | 16.3 | (14.7/6.3) | n.s. | 4.6 | (1.0/1.3) | 0.000 | 15.3 | (3.2/4.7) | n.s. | 17.3 | (5.6/4.5) | 0.045 |
| | f | 3.2 | (1.0/0.4) | | 25.2 | (4.2/11.4) | | 9.9 | (4.3/3.4) | | 16.2 | (2.7/4.8) | | 11.9 | (3.3/2.2) | |
| 10° (25%) | m | 2.4 | (0.7/0.5) | 0.000 | 16.5 | (20.6/3.1) | n.s. | 6.7 | (2.3/1.9) | 0.000 | 18.0 | (3.4/5.5) | n.s. | 18.3 | (8.7/5.4) | 0.009 |
| | f | 4.2 | (1.0/0.7) | | 27.8 | (9.0/9.0) | | 14.6 | (4.3/3.9) | | 13.1 | (3.3/1.4) | | 10.2 | (4.1/3.2) | |
| 20° (48%) | m | 4.2 | (0.8/1.2) | 0.000 | 27.5 | (23.4/12.3) | n.s. | 14.7 | (0.5/4.5) | 0.000 | 23.9 | (1.9/6.0) | n.s. | 17.3 | (9.3/1.6) | n.s. |
| | f | 8.3 | (3.6/1.8) | | 37.5 | (20.9/14.0) | | 24.9 | (7.6/7.4) | | 20.6 | (4.8/2.1) | | 14.8 | (5.0/4.4) | |
| 30° (71%) | m | 6.7 | (1.0/2.0) | 0.000 | 51.3 | (19.4/27.4) | n.s. | 20.3 | (5.8/5.2) | 0.003 | 33.8 | (4.2/7.9) | n.s. | 23.7 | (8.5/6.6) | n.s. |
| | f | 15.8 | (3.7/5.5) | | 50.7 | (29.5/19.2) | | 33.1 | (11/6.1) | | 30.2 | (8.7/5.2) | | 19.7 | (6.0/3.8) | |
| 45° (100%) | m | 20.6 | (14.9/12.3) | n.s. | 69.1 | (41.7/34.2) | n.s. | 36.0 | (7.9/13.6) | 0.001 | 48.6 | (4.0/5.3) | n.s. | 41.2 | (4.4/14.4) | n.s. |
| | f | 29.2 | (9.7/5.4) | | 95.6 | (17.5/40.4) | | 52.5 | (10.5/8.1) | | 47.7 | (8.1/8.7) | | 36.2 | (8.9/9.9) | |

**Opposite force direction**

**forward/backward tilt directions**

Relevant RMS amplitude levels could be observed for OI and MF muscles whereas all other muscles remained at low to negligible RMS amplitudes. Except for OI women showed higher RMS amplitudes for all muscles if compared to men (table 3). This could be proven statistically for RA (10°-60° tilt angles), OE (5°-45° tilt angles), MF (5° - 90° tilt angles), and ES (45° and 60° tilt angles).

Consequently, relative amplitude levels were significantly higher for OE (5° - 45° tilt angles), MF (5° - 90° tit angles), and ES (20° - 90° tilt angles) also. Although men appeared to show higher relative amplitudes for RA and OI too, no systematic effect could be proven.

**sideward tilt directions**

Except for OE, RMS amplitude levels again remained at low to negligible levels throughout the whole tilt range. Slight, but significant differences could be observed for RA (10° - 30° tilt angles), OE (5° and 10° tilt angles), and MF (5°, 10° and 30° tilt angles) with higher RMS levels in the women's group (table 3). Although relative amplitude levels, except RA, appeared to be higher in women, significant differences could be detected only for MF (5° - 45° tilt angles).

For most muscles there was no relationship between relative amplitude and relative torque level. Only for RA and OE a dependency of relative amplitude from torque angle could be observed with increasing relative amplitudes at higher torque levels.

## Discussion

### SEMG normalization

Since SEMG is characterized by high inter-individual differences, absolute RMS amplitudes are difficult to interpret. To minimize the variability of individual RMS amplitude levels data have to be normalized. The golden standard to do this is still the use of maximum voluntary contraction (MVC) levels. But the application of MVC measurements is difficult and calculations are influenced unreliably: familiarity (Mannion et al. 2001), motivation (McNair et al. 1996) and fear of pain (Elfving et al. 2003) are only few examples. The awareness that MVC does not provide consistency has been resolved in investigations by using well defined

Table 3: SEMG amplitudes (Median, (upper/lower quartile range) during whole body tilt in opposite force directions.

| tilt angle (rel. torque) | | RA | | p | OI | | p | OE | | p | MF | | p | ES | | p |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| forward/backward tilt | | | | | | | | | | | | | | | | |
| 5° (9%) | m | 2.1 | (0.5/0.2) | n.s. | 13.5 | (15.1/4.6) | n.s. | 2.6 | (1.8/0.3) | 0.001 | 5.0 | (1.1/1.7) | 0.034 | 8.8 | (6.3/3.8) | n.s. |
| | f | 2.7 | (0.6/0.4) | | 16.4 | (15.3/6.5) | | 4.8 | (2.4/0.7) | | 7.9 | (2.5/1.9) | | 6.2 | (3.4/3.2) | |
| 10° (17%) | m | 2.3 | (0.6/0.4) | 0.019 | 14.8 | (18.9/5.1) | n.s. | 3.1 | (2.1/0.5) | 0.000 | 3.4 | (2.5/0.4) | 0.020 | 6.5 | (4.1/2.7) | n.s. |
| | f | 2.9 | (0.7/0.3) | | 16.8 | (15.6/4.8) | | 6.1 | (1.8/1.0) | | 9.2 | (3.2/3.9) | | 7.6 | (3.3/3.2) | |
| 20° (34%) | m | 2.5 | (0.4/0.4) | 0.021 | 18.5 | (15.4/6.1) | n.s. | 3.5 | (1.5/0.8) | 0.002 | 5.7 | (3.5/1.2) | 0.002 | 5.8 | (3.4/2.8) | n.s. |
| | f | 3.1 | (0.4/0.4) | | 21.5 | (16.0/9.1) | | 6.0 | (1.5/1.4) | | 11.2 | (2.2/2.2) | | 7.9 | (1.8/2.3) | |
| 30° (50%) | m | 2.7 | (0.6/0.4) | 0.021 | 16.9 | (21.7/3.9) | n.s. | 3.7 | (1.3/0.6) | 0.001 | 7.4 | (2/1.8.0) | 0.000 | 6.1 | (4.1/1.2) | n.s. |
| | f | 3.2 | (0.8/0.4) | | 22.1 | (9.6/8.5) | | 6.6 | (3.4/1.1) | | 14.5 | (1.0/1.9) | | 9.3 | (1.6/1.1) | |
| 45° (71%) | m | 3.0 | (0.8/0.2) | 0.019 | 22.1 | (15.4/10.4) | n.s. | 4.5 | (0.8/0.7) | 0.006 | 9.0 | (3.2/2.6) | 0.002 | 6.7 | (0.7/2.1) | 0.006 |
| | f | 3.6 | (0.8/0.1) | | 20.5 | (9.2/7.7) | | 7.3 | (1.4/1.3) | | 15.3 | (1.9/5.0) | | 10.4 | (1.9/2.3) | |
| 60° (87%) | m | 3.5 | (0.4/0.3) | 0.014 | 21.4 | (15.2/9.7) | n.s. | 5.5 | (1.8/1.0) | n.s. | 8.5 | (2.4/2.2) | 0.002 | 6.5 | (0.6/1.0) | 0.002 |
| | f | 4.1 | (0.7/0.2) | | 23.7 | (10.2/9.2) | | 7.0 | (1.8/1.2) | | 12.8 | (2.5/2.2) | | 8.4 | (2.6/0.7) | |
| 90° (100%) | m | 4.5 | (1.5/0.8) | n.s. | 13.9 | (17.0/2.3) | n.s. | 6.8 | (1.7/1.5) | n.s. | 10.0 | (3.8/2.0) | 0.006 | 8.2 | (2.0/1.3) | n.s. |
| | f | 4.7 | (0.7/0.7) | | 19.8 | (7.6/8.2) | | 7.4 | (1.8/0.8) | | 14.7 | (3.7/1.8) | | 9.7 | (2.6/0.6) | |
| sideward tilt | | | | | | | | | | | | | | | | |
| 5° (12%) | m | 2.2 | (0.4/0.5) | n.s. | 8.9 | (15.5/2.0) | n.s. | 3.4 | (0.3/0.9) | 0.000 | 4.5 | (3.8/0.5) | 0.066 | 6.8 | (0.9/1.8) | n.s. |
| | f | 2.6 | (0.6/0.4) | | 11.9 | (12.7/4.7) | | 6.0 | (1.2/1.2) | | 7.5 | (5.0/1.4) | | 8.0 | (1.5/2.7) | |
| 10° (25%) | m | 2.1 | (0.4/0.3) | 0.015 | 9.4 | (9.5/2.7) | n.s. | 4.1 | (0.9/1.3) | 0.001 | 4.7 | (1.0/1.8) | 0.004 | 5.4 | (0.9/1.4) | n.s. |
| | f | 2.6 | (0.7/0.2) | | 10.1 | (9.0/2.4) | | 6.5 | (1.8/1.4) | | 7.6 | (2.9/2.1) | | 4.7 | (2.1/1.1) | |
| 20° (48%) | m | 2.4 | (0.6/0.1) | 0.001 | 4.3 | (3.5/1.2) | n.s. | 7.6 | (1.1/3.0) | n.s. | 5.1 | (1.2/0.7) | n.s. | 5.8 | (1.1/0.6) | n.s. |
| | f | 3.5 | (0.6/0.5) | | 8.4 | (7.2/2.7) | | 9.8 | (1.2/3.4) | | 6.9 | (2.3/1.9) | | 6.4 | (1.7/1.2) | |
| 30° (71%) | m | 3.5 | (0.2/0.3) | 0.000 | 5.2 | (3.0/1.6) | n.s. | 11.0 | (3.3/1.7) | n.s. | 5.1 | (1.9/0.6) | 0.014 | 6.4 | (1.7/1.5) | n.s. |
| | f | 4.2 | (1.0/0.2) | | 8.9 | (7.5/3.1) | | 12.7 | (3.4/4.0) | | 8.0 | (1.4/1.7) | | 6.9 | (1.4/2.1) | |
| 45° (100%) | m | 7.1 | (2.2/2.2) | n.s. | 6.4 | (5.5/1.2) | n.s. | 19.4 | (6.1/6.2) | n.s. | 7.6 | (3.1/1.0) | n.s. | 7.5 | (1.5/1.5) | n.s. |
| | f | 8.0 | (1.6/1.4) | | 10.0 | (7.8/1.9) | | 17.4 | (3.8/3.9) | | 8.6 | (2.3/0.9) | | 7.9 | (3.8/1.1) | |

submaximal levels to establish normative levels (Mirka 1991). In doing this at least motivational and fatigue related influences can be eliminated. For this investigation the simple compensation of the upper body load against gravitation was used to determine reference levels. Furthermore, this method was used because future applications of the device are aimed at rehabilitation and training for low back patients for whom MVC measurements of all investigated trunk muscles are virtually impossible.
However, this normalization method has to be reviewed carefully because the reference level itself served as a measurement point. To test normalization dependent influences on curve shape RMS amplitudes, relative amplitudes based on relative amplitude differences between individual's 50% load level (30° tilt angle) and their untilted position in the device were calculated (figure 5).

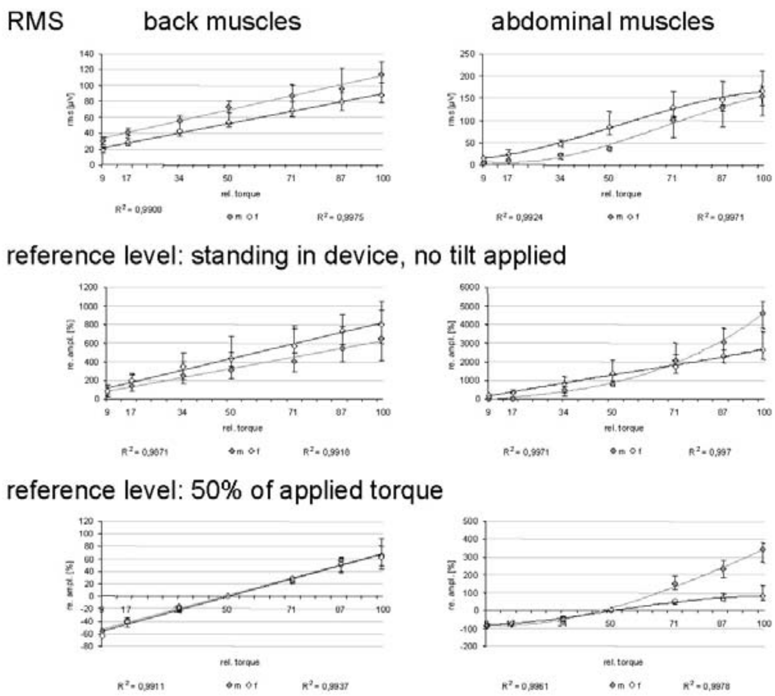


figure 5: Influence of normalization method on curve shape of pooled back and abdominal muscles for main force directions in sagittal plane. Upper row: absolute amplitudes, middle row: data are normalized according SEMG level while subjects were standing in device without any tilt, bottom row: data are normalized according SEMG level while subjects were tilted at 50% of applied torque level (30° tilt angle).

Independent from the method used, curve characteristics remained linear for the pooled back muscles and nonlinear for the pooled abdominal muscles. Calculated abdominal recruitment patterns were subject to normalization. Despite this, RMS curve shape and the maximum normalized data fit nicely, arguing for adequate data treatment.

**Group characteristics**

Anthropometric data of young German adults are available from the literature (Greil 2001). From these data body dimensions and, by derivation, body proportions can be gathered. Although men are 6.8% taller than women (median values men: 180 cm, women: 168 cm), they weigh even 20.3% more than women (men: 74 kg, women: 59 kg (Greil 2001). These data from the literature are in accordance with our group characteristics (see table 1). Lean body mass is assumed to be higher in men as well (men: 84.3%, women: 72.8%), which applies accordingly for muscle mass (Gallagher and Heymsfield 1998; Janssen et al. 2000). However, muscle mass differences are disparate for body regions: the found differences were 40% for the upper body and 33% for the lower body in the Janssen paper (Janssen et al. 2000). Gallagher and Heymsfield found muscle distribution ratios of upper to lower body for 20 year old men to be 0.36 and 0.28 for women of the same age (Gallagher and Heymsfield 1998). This should be reflected in regional body mass proportions also. We found two papers which measured upper body mass (UBM) values. The found gender differences ranged between 18 % (Clark et al. 2003a) and 34 % (Mayer et al. 2002). Also calculation based UBM data are available. For our population, differences in upper body mass between both groups result in 32% (Kankaanpää et al. 1997) or only 24 % (Denner 1998; Zaciorskij et al. 1984). The resulting differences in upper body torque (UBT) at 90° tilt angle are 36 % and 29 %, respectively.

Differences in muscle forces of limb muscles (Lephart et al. 2002) between men and women are obvious: roughly, men's limbs are twice as strong as women's. This applies not entirely for trunk muscles: In the literature we found differences of about 32 % (Clark et al. 2003a) to 40 % (Keller et al. 2004) of maximum extension forces. If data were normalized to body proportions women and men did not differ for extension forces (Keller et al. 2004). For flexion forces a difference of 37 % of normalized force capacity could be proven (Keller et al. 2004)

Furthermore, any possible force production is position related: the more the muscles are stretched the higher the MVC level is (Keller and Roy 2002; Mayer et al. 2002). Summarizing these facts it can be estimated, that during 90° tilt angle both genders reached comparable strain levels during forward flexion. Because of the higher percentage of subcutaneous fat tissue in women one would expect lower amplitudes at comparable strain levels. Data for back tilt situations are subject to larger gender related influences with higher strain levels in women compared with men for corresponding tilt angles. This can be assumed to apply also for sideward tilt directions, but to a lesser extent. Although, RMS amplitudes have to be treated with caution the higher strain level during back tilt seems to be reflected by the higher RMS levels in the women's group.

**Interpretation of data**

In accordance with what could be expected from the literature, if all investigated muscles are regarded, no general difference between men and women was found in our investigation. However, a closer look at muscle groups revealed systematic differences between abdominal and back muscles (table 4). During tilt in the main force directions, women were characterized by higher activation levels of their abdominal muscles, whereas men exhibited higher amplitudes of their back muscles. This could be proven for RMS and relative amplitudes for the abdominal muscles. Relative amplitudes of the back muscles were identical between genders.

Additionally, curve shapes of the amplitude-force relationship for the abdominal muscles differed between genders, but not for the back muscles. In general, abdominal muscles were characterized by a non-linear curve shape and back muscles by a virtually ideal linear amplitude-force relationship. RMS amplitude levels during opposite force directions mostly were higher for women as compared to men.

Table 4: Main tendencies of gender differences of rms values and relative amplitudes during whole body tilt. f: female, m: male, (): only for single tilt angles

| | main force direction | | | | opposite force direction | | | |
|---|---|---|---|---|---|---|---|---|
| | forward/backward | | sideward | | forward/backward | | sideward | |
| | RMS | rel. ampl. | RMS | rel. ampl. | RMS | rel. ampl. | RMS | rel. ampl. |
| RA | f > m | f > m | f > m | f = m | f > m | f = m | (f > m) | f = m |
| OI | f > m | f > m | f = m | f = m | f = m | f = m | f = m | f = m |
| OE | f > m | f > m | f > m | (f > m) | f > m | f > m | (f > m) | (f < m) |
| MF | f < m | f = m | f = m | f = m | f > m | f > m | (f > m) | f > m |
| ES | (f < m) | f = m | (f < m) | f = m | (f > m) | f > m | f = m | f = m |

Since these differences are not only related to muscle groups but also to gender in case of the abdominal muscles muscle specific mechanisms are not adequate to explain this result completely. Therefore, gender related muscular attributes may be responsible for this.

**Morphofunctional background**

Unfortunately we found little information in the literature about fiber type distribution differences between men and women to highlight morphological information. For abdominal muscles we could find only one article by Haggmark and Thorstensson (Haggmark and Thorstensson 1979) who investigated fiber types of all human abdominal muscles. Besides a great inter-individual variability the authors found few predictable differences between the investigated muscles, which were RA, OI, OE and transverse abdominal muscle (TR). In general they found 55-58% Type 1 fibers in all abdominal muscles. Relative cross sectional area (CSA) of Type 1 was 60% for TR and about 54% for the other abdominal muscles (Haggmark and Thorstensson 1979).

For back muscles more data from the literature are available (Mannion 1999; Thorstensson and Carlson 1987), but are also far from complete. In general, although fiber type distribution seems to be comparable between genders, CSA is differently related. The largest fibers are Type 2 for men, but Type 1 for women (Mannion 1999). Therefore, relative CSA of Type 1 fibers are 66,4% for men and 72,8% for women, respectively. In one other study relative CSA of Type 1 fibers was found to be 56% for men and 73% for women, which is different, but does not substantially deviate from the former data (Thorstensson and Carlson 1987).

In general, for both genders the area of Type 1 fibers is less in abdominal muscles as compared to back muscles. This difference should be reflected in electromyographical parameters: Differences are obvious in the amplitude-force relationships of abdominal and back muscles with a non-linear slope for the abdominal muscles (possibly due to the higher proportion of Type 2 fibers) and a linear relationship in the back muscles. These different types of curve shapes are in accordance with different recruitment strategies: if force is increased by increased firing rate the curve is expected to follow the known, virtually linear, force-amplitude relationship (Lawrence and De Luca 1983; Solomonow et al. 1990). In contrast, if force increase is mainly due to increased recruitment of motor units the force-amplitude curve is expected to change its slope towards an s-shaped mode (Solomonow et al. 1990).

Combining the histological and electromyographical facts, fiber composition and recruitment type may be linked. The EMG-force relationship hypothetically may follow these rules: A high proportion of type 1 fibers would primarily be correlated with increased firing rate, a low proportion of type 1 fibers, in contrast, with the recruitment of additional motor units. Up to now no data are available to support this hypothesis.

Since for the back muscles, the relative CSA of Type 1 fibers was found to be much higher in women compared to men, an analogous, aggravating effect is possible, resulting in the observed higher amplitude levels in men. Necessarily, women should demonstrate a higher relative CSA of Type 1 fibers of their abdominal muscles too to prove this hypothesis. As a support of this assumption, we found a less s-shaped, more linear curve for the women if compared with the curves of the investigated men.

Histological data from a large sample are available, but detailed data about the respective parameters of trunk muscles unfortunately were not included (Simoneau and Bouchard 1989). There is need for further detailed investigations.

It has to be assumed, that strain levels were different between genders for backward and also sideward tilt directions with higher strain levels for the group of investigated women. This should evoke additional recruitment of motor units, resulting in more deviation from linearity (Solomonow et al. 1990). This could not be found in our investigation. Therefore, the observed differences in curve shape between genders cannot be explained by the higher strain levels of the investigated women.

**Muscular fatigue**

Muscular fatigue might be considered as another possibility to explain these results. There are hints in the literature, that women tend to be more fatigue resistant than men (Mannion et al. 1997). But fatigue resistance seems to be dependent on contraction type: if persistent contractions were investigated no gender differences could be observed for arm muscles, whereas intermittent contractions evoked higher degrees of fatigue in men (Clark et al. 2003b). Results for back muscles are available again arguing for higher fatigue resistance in women compared to men (Kankaanpää et al. 1998). This even when persistent contractions were investigated. Data for abdominal muscles were not available from the literature.

Since measurement time was restricted to 10 seconds, sufficient breaks were included, and positions were applied randomly, fatigue should not have had a systematic influence on the results.

**Co-contraction**

The constantly higher RMS amplitudes for the women's group during the opposite force directions cannot be explained by gender related muscle compositions, nor anthropometric differences. For this phenomenon gender specific activation strategies have to be considered. Women already showed a higher degree of co-contraction levels for shoulder muscles (Anders et al. 2004). This might argue for higher stability demands in women. Since the investigation was carried out in a new device this could also have behavioral reasons. A high degree of co-contraction is always connected with unfamiliar tasks. The more habituated with a certain task a person is, the more muscles are utilized optimally. This could be proven for timing (Carson and Riek 2001) and co-ordination parameters (Carson and Riek 2001; Lay et al. 2002). Since all subjects were using the device for the first time habituation does not apply for the observed differences.

Anxiety was not reported by the subjects, but holding the unsupported trunk stable approximately 1.5 m over floor level might impress women more than men (Courtenay 1998). This may lead to the observed differences of trunk muscle RMS amplitudes, but cannot be proven with the actual data. Since this investigation was focused on the identification of muscle related activation strategies, intermuscular co-ordination strategies were not analyzed. For this purpose data would have to be normalized according to relative contribution to every single task, e.g. tilt angle separately (van Dieen et al. 2003). Therefore, at present it remains unproven if relevant gender specific intermuscular recruitment pattern of trunk muscles exist.

**Conclusions**

Gender specific differences in amplitude-force relationships of trunk muscles could be identified. Although men showed higher RMS amplitudes of their back muscles the linear amplitude force relationship did not differ between genders. In contrast, women showed higher amplitudes of their abdominal muscles. The non-linear amplitude-force relationship of abdominal muscles was gender related with a less non-linearity in women compared with men.

Gender related differences in fiber composition between abdominal and back muscles in general and between genders are known from the literature, which might explain the results. Co-activation of antagonistic muscles was more pronounced in the women's group, possibly caused by behavioral reasons.

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1. Frau Beatrix Wenzel: Teilnahme an Messungen für Propriomed-Untersuchungen

2. Herr Gunther Brose: Teilnahme an Messungen für Centaur-Untersuchungen

Weitere Personen waren an der inhaltlich-materiellen Erstellung der Arbeit nicht beteiligt. Insbesondere habe ich hierfür nicht die entgeltliche Hilfe von Vermittlungs- bzw. Beratungsdiensten in Anspruch genommen. Niemand hat von mir unmittelbar oder mittelbar geldwerte Leistungen für Arbeiten erhalten, die im Zusammenhang mit dem Inhalt der vorgelegten Arbeit stehen.

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