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Biomechanical comparison of walking with a new, wearable rehabilitation training device to Nordic walking and regular walking in people with chronic low back pain

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Abstract

Physical activity, particularly walking, is commonly used for the treatment of diseases such as low back pain. In this study, the effects of walking wearing the new ToneFit Reha training belt (TFR) were compared to both Nordic walking and regular walking. The TFR is intended to intensify the effects of walking through the integration of two adjustable resistance handles.

Ten patients with low back pain performed regular walking, Nordic walking and walking with the TFR in a movement laboratory. The kinematics of the trunk, upper extremities and lower extremities were measured, and the activity of the trunk and upper extremity muscles recorded. Data were analysed by repeated-measures ANOVA and paired t-test.

Kinematics indicated that walking with the TFR introduces instability that was mitigated by a delayed peak trunk rotation (peak at 63.3% gait cycle, vs. 52.8% in walking ($p=0.001$) and 51.0% in NW ($p=0.007$)). Upper extremity kinematics (constrained elbow flexion, high peak shoulder abduction) showed movement patterns that need to be considered when training over a longer period. Increased muscle activity was observed especially for upper extremity muscles, when training with TFR. Overall, walking with the TFR was found to be a suitable therapy for use in a rehabilitation setting.

Keywords: wearable training device; low back pain; unstable walking

Word count: 3362

Introduction

In the field of physiotherapy, treatment practice has been systematically investigated for many years, resulting in a shift from mainly passive to active therapy. Physical activity is now predominantly recommended for the treatment of non-communicable diseases, such as musculoskeletal disorders [1]. One musculoskeletal disorder is low back pain (LBP), which is the worldwide leading cause for impairment [2]. As a therapy approach for LBP, walking with poles, called Nordic walking (NW), has been the subject of scientific investigation. It has been shown to increase core stability [3], as well as affecting the resting heart rate, blood pressure, exercise capacity, maximal oxygen consumption, and quality of life beneficially. These benefits have also been found to apply to patients with other diseases, such as obesity, coronary heart disease, pain in general, breast cancer, and Parkinson's disease [4]. NW was found to have a more intense workout effect with less perceived exertion [5].

Besides NW poles, other products are also used to enhance the effects of walking or running. These include carrying weight pockets while jogging or wearing a belt with integrated sling trainers. Another recently developed fitness training product is the ToneFit belt, a device that is worn around the waist. It has been designed to mimic and intensify the beneficial movement patterns of regular walking or jogging. This is achieved through two independent pull-push elements on either side of the belt, with non-adjustable resistance, which raise the intensity of walking and increase the effects of training on strength and endurance, in particular on the upper extremities and trunk. It is assumed that walking with the ToneFit causes instability, which is compensated by increased activity of the trunk musculature. Preliminary, unpublished studies of the ToneFit did not analyze movement patterns, but they did indicate, based on heart rate and oxygen consumption, that the product offers a more intense training than NW [6].

A suitable therapy for LBP patients could include the use of such a training device because LBP has been shown to change the stability of the trunk musculature [7–9]. Also, it was observed that LBP patients can be subdivided into two groups: some limit their movement, while the others avoid high muscle activation [10]. Both could profit from enhancing the movement and increasing trunk muscular activity. However, the currently available products do not allow the pull-push element resistance to be adjusted to suit the condition of individual patients. The ToneFit has therefore been further developed to include two independent pull-push elements on either side of the belt with adjustable resistance, to allow for individual adjustment. This revised product is called the ToneFit Reha (TFR). Through this modification, it is thought that the TFR could be a suitable device for rehabilitation purposes, by increasing the training intensity of the upper extremities and trunk. It is unclear, however, whether the movement pattern using the TFR lies within a satisfactory physiological range, or whether it restricts the range of motion, which could lead to incidence of pain [11]. This study was designed to compare the biomechanical effects on LBP patients of regular walking, NW and walking with the TFR. The physiological kinematics and the effects on muscular activity of walking using the TFR were investigated. It was observed whether TFR shows another kinematic pattern and if muscle activity can be increased by walking with TFR.

Methods

Design

This study was designed as a cross-sectional, monocentric pilot study.

Study population

Ten LBP patients were recruited through a mailing at the university and through physiotherapy practices in the Winterthur area. Inclusion criteria were reported pain of at least three months during the last year and a minimum of five points on the Oswestry Disability Index [12]. Patients must have had no diagnosis of scoliosis or spine stiffening. Patients with health conditions that impair TFR walking (e.g. shoulder pain) were excluded. All patients signed an informed consent form prior to the start of the study and approval was received from the cantonal ethics committee Zurich (Nr. 2018-00821) and Swissmedic (Nr. 2018-MD-0010).

Investigational product

The newly developed TFR product was investigated in this study. The portable training device is worn around the waist (belt, see Figure 1) and consists of two pull-push resistance handles at the left and right sides. During walking, these handles are pushed and pulled by the arm swing. The bidirectional resistance of the handles can be adjusted individually at each side through a smartphone application. Resistance is achieved with active damping using a magnetorheological fluid actuator (MRF Actor) and a low-power control unit (logic and battery), which communicates with a smartphone application (Figure 1: ToneFit Reha worn by a participant while walking

Figure).

Figure 1,2 around here

As control conditions, NW and regular walking were chosen. NW was performed with commercially available Nordic walking poles that were adjustable in height.

Data collection

The data collection took place in the ZHAW movement laboratory. Kinematic data was captured using an opto-electronic motion capturing system at 240 Hz (Vicon Vantage with Nexus 2, Vicon Motion Systems Ltd, UK) with 12 infrared cameras and reflective markers. Electromyographic data (EMG) was captured by surface electrodes with a wireless transmitter at 1200 Hz (Myon AG, Switzerland). Walking speed was measured using two light barriers placed at a known distance (MicroGate, Bozen, Italy).

After instruction and familiarization with NW and walking with the TFR, the patients were prepared with the markers and electrodes. The skin was prepared by shaving and disinfecting with alcohol before electrode placement. Electrodes were placed according to SENIAM [13] on the left side and right side muscles for M. biceps brachii, M. triceps brachii, M. erector spinae, M. multifidi, M. obliquus externus and M. pectoralis major. Markers were placed on the upper and lower extremities, as well as on the spine, according to the marker models described in the next section. Once the markers and electrodes were attached, the patients were instructed to walk in a figure of eight form to guarantee continuous walking [14]. They first performed regular overground walking (walking) at a self-selected speed, followed by NW and, finally, the TFR walking. For NW and TFR, the same speed as for walking needed to be maintained (±5%). For each condition, a total of 10 gait cycles were captured. The patients were continuously walking in the figure of eight form in the movement laboratory (total length of 10 meters) until the 10 gait cycles were recorded.

Before starting the measurement, patients were asked to report their pain of the last 24 hours on a numerical rating scale from 0 (no pain) to 10 (strongest pain imaginable). After each condition, the pain rating was repeated. In case of more than 3 points increase to the last 24 hours, the measurement was stopped.

Marker Models

To determine the kinematics of the lower extremities, a previously developed cluster marker set with functional determination of joint centers for the hip and knee joint was used [15]. Trunk angles were calculated based on a spine model [16]. For the upper extremities, a similar cluster marker model was developed, which is described in detail in the appendix.

Data analysis

Kinematics and EMG data were analyzed for the left and right steps and for their ipsilateral segments and muscles. Of the 10 measured gait cycles of each participant and task, the mean was calculated und used for further analysis.

Processing of kinematics was performed in Matlab R2019a. Marker data was filtered with a 4th order Butterworth filter and a cut-off frequency of 7Hz. For kinematics, the variables peak angle and range of motion (ROM) were determined.

EMG signals were filtered with a 2nd order Butterworth filter, with a cut-off frequency of 10Hz for high-pass and 500Hz for low-pass filtering. The root mean square (RMS) window was 100ms. A static trial was performed in an upright stance with relaxed muscles, from which the activation threshold was determined for each muscle by the mean signal plus k-times the standard deviation (k=3 for trunk, k=10 for upper extremities). All EMG data were expressed as relative signals to those of regular walking. The duration of muscle activity was defined as the time that the muscle was active, meaning above the threshold. The mean activity was only calculated for the time the muscle was active, while maximal activities were calculated over the whole trial.

For all discrete variables, the differences of TFR walking to NW and regular walking were of interest. Therefore, statistical analysis was calculated with repeated measures ANOVA ($p<0.05$). For significant differences, a paired t-test with Bonferroni correction was performed. All statistical analysis were performed in Matlab R2019 (MathWorks Inc., USA).

Results

Ten patients with low back pain (LBP) were recruited (age 42.1 ± 5.4 years, mass 71.4 ± 8.0 kg, height 172.1 ± 5.2 cm, Oswestry Disability Index 9.2± 3.4). Their average walking speed was 1.51 ± 0.08 m/s. No measurement had to be stopped because of the pain rating.

Kinematics

Trunk

For all three conditions, the trunk showed a rotation in the transverse plane to the ipsilateral side during the heel-strike, followed by a rotation to the contralateral side during the mid-gait cycle (Figure). The movements in the other planes (lateral flexion and flexion) showed less ROM compared to rotation (Table 1). Rotation ROM was between 19.3° and 23.5°.

For trunk ROM, no significant differences between the three conditions were found. However, the timing of the rotation differed significantly (Table 1). Peak rotation occurred later in the gait cycle for TFR walking than for NW and regular walking. This means that the trunk stayed in a more contralateral rotation during push-off (Figure).

Figure 3 and Table 1 around here

Upper extremities

In the wrist, differences between the conditions are visible for ulnar deviation (

Figure). NW has the largest ROM, followed by TFR walking and regular walking. On the left side, TFR walking has a larger pronation ROM than the other conditions (Table 2).

For ulnar deviation, the maximum is also significantly different in TFR walking to the other conditions. While regular walking and NW lead to ulnar deviation, TFR walking shows a wrist movement towards radial deviation, especially during the swing phase.

Figure 4 around here

For the elbow, NW and TFR walking showed a larger maximal flexion (Figure 5,

Table 3). While NW shows a large ROM, the elbow in TFR walking stays in the flexed position with a relatively low ROM. Moreover, with TFR walking the elbow is more adducted.

Figure 5 around here

The shoulder was observed to be in a more abducted position in TFR walking (Figure), while the ROM did not differ between conditions.

Figure 6, Table 2, Table 3 around here

Lower extremities

The kinematics of the lower extremities differed only slightly between the three conditions. Some significant differences were found for ROM measures between TFR walking and regular walking for left ankle inversion (TFR 12.5° (2.8) vs Walking 10.7° (3.1)) and left knee adduction (TFR 12.7° (3.5) vs Walking 11.6° (3.1)) (Figure 7). For the ankle, the inversion is increased at the end of the stance phase as well as at the end of the swing phase. For the knee, more abduction is found in TFR during the swing phase.

Figure 7 around here

EMG

For EMG data analysis, one patient had to be excluded due to an erroneous sampling frequency, making a comparison to the other patients impossible. Therefore, only nine patients were included in this analysis. All data can be found in Table 4 and

Figure .

Core muscles

The muscle activity of M. multifidi, M. erector spinae, M. obliquus externus and M. pectoralis were measured for both the left and right sides. M. pectoralis left had to be excluded because the signal artefacts of the heart muscle were too dominant. M. pectoralis right was less affected by the artefacts and was therefore included. When walking with TFR, a high variance (represented by the standard deviation) in M. pectoralis muscle activity was seen. A longer onset duration than for regular walking was shown.

For M. multifidi, the mean and the maximum activity of the right side was increased in TFR walking compared to NW. The M. erector spinae showed a higher maximal activity for the left part during TFR walking compared to regular walking. In addition, a longer onset was observed for both sides.

For M. obliquus externus, no significant changes in muscle activity while using the TFR were observed.

Upper extremities muscles

During regular walking, the upper extremity muscles worked at a low level; for TFR walking a significant increase in mean activity compared to regular walking was observed. Also, the maximal activity increased in TFR walking compared to regular walking, except for the M. biceps right. However, for M. triceps right an even higher mean and maximum activity was found during NW than TFR walking. The activity duration of all upper extremity muscles was increased with TFR walking compared to regular walking; between TFR walking and NW no significant changes were observed.

For upper extremity muscles, higher standard deviations than for core muscles were observed, e.g., M. triceps brachii in NW with up to 2000%. Also, a high inter-patient variability was seen.

Figure 8, Table 4 around here

Discussion

The aim of this study was to investigate the biomechanics and muscle activity effects of walking with the ToneFit Reha belt (TFR) compared to NW and regular walking.

During walking with the TFR, trunk ROM did not differ from regular walking and NW. In LBP therapy, it is crucial to not limit the range of motion, as it may be already reduced in some patients [10,17]. This implies, that concerning this aspect TFR is suitable for LBP, as it allows trunk movement in the same range as regular walking and NW. However, the trunk was assumed to be affected by a change in stability. The pull and push movement with the handles had to be counterbalanced by trunk kinematics and muscle activity. It was observed that while walking with the TFR, patients showed a late peak trunk rotation angle. This phenomenon has not been described in existing literature; it could be induced by a sense of instability while walking with the TFR. It seems possible that the challenging task of changing the trunk and arm movement direction requires greater stability when walking with the TFR than during regular walking and is therefore delayed to the more stable double support phase. In the double support phase, unstable movements, such as the change of trunk rotation direction, require less trunk muscle activity compared to single leg support and can be performed with less effort. This explanation was also supported by the data on core muscle activity. The activation of core muscles was thought to differ during walking with the TFR, since they would be necessary to counterbalance the instability caused by the belt. However, this effect was observed only partially, and it can be assumed that the instability was rather compensated by kinematics than muscle activity. The use of stability aspects during general exercises for rehabilitation has been shown to be beneficial. In LBP patients performing stability exercises, lumbar instability was shown to be improved [18], the incidence of pain reduced, and protection from injury raised [19]. It may therefore be assumed that the instability caused by TFR walking is beneficial for rehabilitation purposes.

For the upper extremity muscles, a high inter-patient variability was found, which could be due to the different levels of experience of patients walking with NW poles and to varying adaptation to TFR walking. Although patients were given time to familiarize themselves with NW and TFR, their walking techniques cannot be compared to that of experienced users. However, the results imply that the findings of increased muscle activity during NW compared to regular walking [20–23] can be transferred to TFR walking. Further investigation with a larger population is needed to confirm a more general outcome.

Kinematics of the wrist, elbow and shoulder changed in TFR walking. Even though increasing ROM is thought to be beneficial in LBP, the risk for other musculoskeletal disorders with non-physiological movement patterns need to be considered, especially in upper and lower extremities that are not directly linked to LBP. Earlier studies have found an increased risk of wrist musculoskeletal disorders, with ulnar deviations above 20° and wrist extensions above 15-20° [24–26]. Walking with the TFR caused increased radial deviation, which is less associated with pain than an increased ulnar deviation. No increased extension was found. Additionally, a constrained elbow flexion with TFR walking was observed, which might lead to pain [11,24]. However, another study has shown that a minimum loading of two hours daily was the critical time for constrained work to become harmful [27]. The loading time should be considered when using the TFR. As well as the elbow, the shoulder is abducted while walking with the TFR. This constantly abducted shoulder position could also induce pain [25]. Future studies should consider the constrained elbow angle and abducted shoulder, to examine if this position leads to pain, when training over a longer duration. For the lower extremities only small differences between TFR walking and regular walking were found. Increased ankle inversion and knee abduction are risk factors for injuries [28,29]. As the increased knee abduction occurred during swing phase, the impact is lower, than if it had been during stance. Also, total ab-/adduction ROM differed around 1°, which can therefore be assumed as non-clinically relevant. Both ankle inversion and knee adduction ROM were only increased on the left side. Therefore, no clear indication for injuries is available, but especially ankle inversion should be taken into consideration for future studies. Overall, regarding all angles of the lower extremities, we conclude that walking with the TFR did not change movement patterns in the lower extremities except for ankle inversion.

LBP patients are known to show asymmetries in their left/right side behaviour [30–34]. Therefore, parameters were calculated for both sides separately. It was observed that the significant differences between conditions were equal for left and right trunk kinematics but deviated for left and right sides in upper and lower extremities kinematics and muscular activity.

There are some limitations to this study. Measurement of muscle activity was limited to the surface muscles. Deep muscle activity could also have been changed during TFR walking since they play an important role in stabilizing the trunk. This effect, however, was not measurable with the chosen non-invasive approach. Also, the small sample size is a limitation, which does not allow generalizable results. In addition, the order of conditions was non-randomized. This was chosen, because the speed of regular walking was used for the other two walking tasks. Furthermore, the patients were inexperienced in walking with the TFR, resulting in large inter-patient variability in their reactions. Finally, it remains unknown as to how prolonged training with the TFR would affect the walking pattern. Especially to the changes in elbow and shoulder kinematics it should be paid attention in future studies, once the TFR is used over a longer duration.

Conclusion

While the biomechanics of the lower extremities remained almost unchanged, upper extremities and the trunk displayed changed movement patterns when walking with the TFR compared to regular walking. Some of the study findings indicate an alteration in balance when walking with the TFR, which appears to be compensated by kinematics. It is assumed that the observed movement does not increase the risk of disorders, but it is necessary to consider the length of time exposed to the restricted postures. To minimize the risk of pain through exposure to the restricted movement of the upper extremities, it may be necessary to include recommendations for preventative measures (e.g., stretching exercises) in the product information.

The trunk muscles showed no consistent pattern of increased activity. However, a tendency towards increased back muscle activity, with no change to the abdominal muscles, was identified. The upper extremities showed the expected increase in muscle activity when walking with the TFR compared to regular walking, but these effects were similar to Nordic walking.

In conclusion, the TFR has been found to be a suitable device for use in a rehabilitation setting, but further research into the detailed instructions and training with the TFR, as well as the duration of the training, is required.

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Declaration of interest

The authors declare no conflict of interests.

References

[1] Bloxham S, Barter P, Scragg S, et al. Person-Centered, Physical Activity for Patients with Low Back Pain: Piloting Service Delivery. Healthcare. 2016;4:28.

[2] Hoy D, March L, Brooks P, et al. The global burden of low back pain: estimates from the Global Burden of Disease 2010 study. Ann Rheum Dis. 2014;73:968–974.

[3] Zoffoli L, Lucertini F, Federici A, et al. Trunk muscles activation during pole walking vs. walking performed at different speeds and grades. Gait Posture. 2016;46:57–62.

[4] Tschentscher M, Niederseer D, Niebauer J. Health Benefits of Nordic Walking. Am J Prev Med. 2013;44:76–84.

[5] Church TS, Earnest CP, Morss GM. Field Testing of Physiological Responses Associated with Nordic Walking. Res Q Exerc Sport. 2002;73:296–300.

[6] Schrag R. tone-fit: Evaluierung eines Lauftrainingsgürtels [Unpublished bachelor thesis]. Berner Fachhochschule; 2013.

[7] Ghamkhar L, Kahlaee AH. Trunk Muscles Activation Pattern During Walking in Subjects With and Without Chronic Low Back Pain: A Systematic Review. PM&R. 2015;7:519–526.

[8] Hanada EY, Johnson M, Hubley-Kozey C. A Comparison of Trunk Muscle Activation Amplitudes During Gait in Older Adults With and Without Chronic Low Back Pain. PM&R. 2011;3:920–928.

[9] Ranger TA, Cicuttini FM, Jensen TS, et al. Are the size and composition of the paraspinal muscles associated with low back pain? A systematic review. Spine J. 2017;17:1729–1748.

[10] van Dieën JH, Reeves NP, Kawchuk G, et al. Motor Control Changes in Low Back Pain: Divergence in Presentations and Mechanisms. J Orthop Sports Phys Ther. 2019;49:370–379.

[11] Nordander C, Ohlsson K, Åkesson I, et al. Risk of musculoskeletal disorders among females and males in repetitive/constrained work. Ergonomics. 2009;52:1226–1239.

[12] Mannion AF, Junge A, Fairbank JCT, et al. Development of a German version of the Oswestry Disability Index. Part 1: cross-cultural adaptation, reliability, and validity. Eur Spine J. 2006;15:55–65.

[13] Hermens HJ, Commission of the European Communities, Biomedical and Health Research Programme, editors. The state of the art on signal processing methods for surface electromyography. 1999.

[14] König N, Singh NB, von Beckerath J, et al. Is gait variability reliable? An assessment of spatio-temporal parameters of gait variability during continuous overground walking. Gait Posture. 2014;39:615–617.

[15] List R, Gülay T, Stoop M, et al. Kinematics of the Trunk and the Lower Extremities During Restricted and Unrestricted Squats: J Strength Cond Res. 2013;27:1529–1538.

[16] Rast FM, Graf ES, Meichtry A, et al. Between-day reliability of three-dimensional motion analysis of the trunk: A comparison of marker based protocols. J Biomech. 2016;49:807–811.

[17] Tsigkanos C, Demestiha T, Spiliopoulou C, et al. Gait kinematics in Low Back Pain: A non-linear approach. J Back Musculoskelet Rehabil. 2021;34:707–714.

[18] Javadian Y, Akbari M, Talebi G, et al. Influence of core stability exercise on lumbar vertebral instability in patients presented with chronic low back pain: A randomized clinical trial. Casp J Intern Med. 2015;6:98–102.

[19] Behm DG, Drinkwater EJ, Willardson JM, et al. Canadian Society for Exercise Physiology position stand: The use of instability to train the core in athletic and nonathletic conditioning. Appl Physiol Nutr Metab. 2010;35:109–112.

[20] Boccia G, Zoppirolli C, Bortolan L, et al. Shared and task-specific muscle synergies of Nordic walking and conventional walking. Scand J Med Sci Sports. 2018;28:905–918.

[21] Pellegrini B, Peyré-Tartaruga LA, Zoppirolli C, et al. Exploring Muscle Activation during Nordic Walking: A Comparison between Conventional and Uphill Walking. PLOS ONE. 2015;10:e0138906.

[22] Schiffer T, Knicker A, Montanarella M, et al. Mechanical and physiological effects of varying pole weights during Nordic walking compared to walking. Eur J Appl Physiol. 2011;111:1121–1126.

[23] Shim J, Kwon H, Kim H, et al. Comparison of the Effects of Walking with and without Nordic Pole on Upper Extremity and Lower Extremity Muscle Activation. J Phys Ther Sci. 2013;25:1553–1556.

[24] Kaufman-Cohen Y, Portnoy S, Sopher R, et al. The correlation between upper extremity musculoskeletal symptoms and joint kinematics, playing habits and hand span during playing among piano students. PLoS ONE [Internet]. 2018 [cited 2019 Jun 25];13. Available from: https://www.ncbi.nlm.nih.gov/pmc/articles/PMC6300245/.

[25] Tittiranonda P, Burastero S, Rempel D. Risk factors for musculoskeletal disorders among computer users. Occup Med Phila Pa. 1999;14:17–38, iii.

[26] Wahlström J. Ergonomics, musculoskeletal disorders and computer work. Occup Med. 2005;55:168–176.

[27] Herquelot E, Bodin J, Roquelaure Y, et al. Work-related risk factors for lateral epicondylitis and other cause of elbow pain in the working population. Am J Ind Med. 2013;56:400–409.

[28] Wilczyński B, Zorena K, Ślęzak D. Dynamic Knee Valgus in Single-Leg Movement Tasks. Potentially Modifiable Factors and Exercise Training Options. A Literature Review. Int J Environ Res Public Health. 2020;17:8208.

[29] Kobayashi T, Tanaka M, Shida M. Intrinsic Risk Factors of Lateral Ankle Sprain: A Systematic Review and Meta-analysis. Sports Health. 2016;8:190–193.

[30] Rahimi A, Arab AM, Nourbakhsh MR, et al. Lower limb kinematics in individuals with chronic low back pain during walking. J Electromyogr Kinesiol Off J Int Soc Electrophysiol Kinesiol. 2020;51:102404.

[31] Al-Eisa E, Egan D, Wassersug R. Fluctuating asymmetry and low back pain. Evol Hum Behav. 2004;25:31–37.

[32] Hao Z, Xie L, Wang J, et al. Spatial Distribution and Asymmetry of Surface Electromyography on Lumbar Muscles of Soldiers with Chronic Low Back Pain. Pain Res Manag. 2020;2020:6946294.

[33] Fortin M, Rizk A, Frenette S, et al. Ultrasonography of multifidus muscle morphology and function in ice hockey players with and without low back pain. Phys Ther Sport Off J Assoc Chart Physiother Sports Med. 2019;37:77–85.

[34] Van Dillen LR, Bloom NJ, Gombatto SP, et al. Hip rotation range of motion in people with and without low back pain who participate in rotation-related sports. Phys Ther Sport. 2008;9:72–81.

Tables with captions

Table 1: Mean (± standard deviation) of ROM [°] and time of maximal angle [% gait cycle] of the trunk angles (n=10); * sign. difference between TFR and NW; † sign. difference between TFR and Walking; in brackets p-value for significant results

		Trunk during left step				Trunk during right step			
		TFR	**NW**	**Walking**	**Sig.**	**TFR**	**NW**	**Walking**	**Sig.**
ROM [°]	**rotation**	23.5 (7.2)	19.3 (7.4)	21.6 (7.6)	-	23.3 (6.9)	19.3 (7.2)	21.6 (7.6)	-
	lat. flexion	13.2 (5.1)	13.2 (4.9)	14.6 (4.5)	-	13.2 (5.1)	13.0 (4.9)	14.6 (4.7)	-
	flexion	5.6 (2.2)	5.6 (1.3)	5.3 (1.4)	-	5.7 (2.4)	5.5 (1.2)	5.1 (1.3)	-
Time of peak [% gait cycle]	**rotation**	63.3 (8.6)	51.0 (3.7)	52.8 (3.1)	* (p=0.007) † (p=0.001)	63.8 (7.7)	51.6 (4.4)	51.8 (4.7)	*, † (p<0.001)

Table 2: Mean (± standard deviation) range of motion (ROM) of the upper extremities (n=10); * sign. difference between TFR and NW; † sign. difference between TFR and Walking; in brackets p-value for significant results

ROM [°]		Left side				Right side			
		TFR	NW	Walking	Sig.	TFR	NW	Walking	Sig.
Wrist	Palmar flexion	25.8 (6.1)	23.1 (11.7)	13.9 (5.1)	† (p<0.001)	22.3 (6.5)	20.8 (8.5)	17.2 (6.0)	-
	Ulnar deviation	12.8 (3.9)	29.0 (9.3)	11.7 (6.1)	* (p<0.001)	18.4 (5.7)	25.3 (7.3)	11.1 (5.1)	*† (p=0.01)
	Pronation	20.6 (6.6)	14.0 (7.8)	14.7 (7.5)	* (p=0.006) †(p<0.005)	15.6 (4.1)	15.5 (9.5)	11.7 (4.9)	-
Elbow	Flexion	18.9 (8.4)	53.3 (17.8)	44.0 (12.1)	*† (p<0.001)	21.0 (10.1)	57.4 (16.1)	37.1 (12.0)	*† (p<0.001)
	Abduction	12.9 (3.3)	11.2 (3.5)	10.3 (5.4)	-	12.9 (5.3)	12.3 (4.0)	8.0 (2.7)	† (p=0.007)
	External rotation	9.2 (3.1)	15.3 (5.4)	13.7 (5.4)	* (p<0.001)	8.7 (3.1)	15.8 (5.8)	13.3 (4.8)	* (p<0.001) † (p=0.014)
Shoulder	Flexion	31.7 (6.4)	25.4 (10.0)	32.3 (9.1)	-	40.0 (8.6)	25.4 (7.0)	34.9 (17.2)	* (p<0.001)
	Abduction	9.7 (2.3)	7.3 (3.4)	9.1 (3.1)	-	8.5 (1.5)	7.8 (3.3)	9.6 (3.6)	-
	External rotation	16.6 (5.4)	13.2 (5.6)	17.3 (7.2)	-	22.8 (7.1)	15.3 (5.7)	18.0 (8.0)	* (p=0.014)

Table 3: Mean (± standard deviation) peak angles of the upper extremities (n=10); * sign. difference between TFR and NW; † sign. difference between TFR and Walking.

Peak angles [°]		Left side				Right side			
		TFR	NW	Walking	Sig.	TFR	NW	Walking	Sig.
Wrist	Ulnar deviation	-2.3 (12.7)	16.0 (10.4)	9.6 (9.9)	* (p<0.001) † (p=0.002)	1.8 (9.3)	14.9 (12.5)	11.7 (11.0)	* (p=0.008) † (p=0.002)
Elbow	Abduction	-9.6 (6.3)	-2.6 (8.5)	-2.6 (5.6)	* (p=0.002) † (p=0.005)	-10.6 (8.6)	-2.2 (7.8)	-5.3 (5.7)	* (p<0.001) † (p=0.007)
Elbow	Flexion	72.9 (11.0)	68.4 (15.3)	40.4 (14.3)	† (p<0.001)	76.8 (12.4)	73.1 (12.9)	39.6 (12.0)	† (p<0.001)
Shoulder	Abduction	18.0 (3.9)	8.1 (3.9)	7.8 (3.5)	*† (p<0.001)	16.0 (3.6)	9.5 (5.1)	9.2 (4.9)	* (p=0.002) † (p=0.002)

Table 4: Mean and maximum muscle activation and duration of activity; Mean (SD); * sign. difference between TFR and NW; † sign. difference between TFR and Walking

EMG Activity					
		TFR	**NW**	**Walking**	**Sig.**
M. multifidi left	**Mean activity [% Ref]**	61.7 (18.3)	57.5 (16.7)	56.5 (13.0)	-
	Max. activity [% Ref]	122.1 (40.2)	105.8 (26.6)	100 (0.0)	-
	Activity duration [s]	0.50 (0.32)	0.49 (0.33)	0.45 (0.32)	-
M. multifidi right	**Mean activity [% Ref]**	69.3 (27.6)	58.4 (21.1)	57.6 (15.8)	* (p=0.005)
	Max. activity [% Ref]	147.1 (66.2)	104.1 (27.0)	100.0 (0.0)	* (p=0.005)
	Activity duration [s]	0.40 (0.18)	0.40 (0.17)	0.34 (0.11)	-
M. erector spinae left	**Mean activity [% Ref]**	44.6 (16.3)	41.7 (15.8)	45.1 (16.7)	-
	Max. activity [% Ref]	106.3 (7.7)	101.8 (14.3)	100.0 (0.0)	† (p=0.013)
	Activity duration [s]	0.57 (0.21)	0.53 (0.22)	0.42 (0.17)	† (p=0.001)
M. erector spinae right	**Mean activity [% Ref]**	42.4 (15.5)	41.0 (14.4)	43.5 (14.3)	-
	Max. activity [% Ref]	98.9 (20.7)	100.8 (27.2)	100.0 (0.0)	-
	Activity duration [s]	0.58 (0.15)	0.60 (0.20)	0.38 (0.13)	† (p<0.001)
M. obliquus externus left	**Mean activity [% Ref]**	78.9 (20.0)	78.9 (11.7)	70.5 (14.7)	-
	Max. activity [% Ref]	126.4 (32.4)	118.4 (20.9)	100.0 (0.0)	-
	Activity duration [s]	0.56 (0.26)	0.63 (0.26)	0.54 (0.24)	-
M. obliquus externus right	**Mean activity [% Ref]**	78.7 (31.6)	83.7 (28.8)	75.2 (27.4)	-
	Max. activity [% Ref]	110.0 (25.6)	121.7 (28.6)	100.0 (0.0)	-
	Activity duration [s]	0.53 (0.28)	0.65 (0.39)	0.47 (0.35)	-
M. pectoralis right	**Mean activity [% Ref]**	171.0 (119.0)	111.0 (15.9)	103.0 (9.3)	-
	Max. activity [% Ref]	313.5 (270.4)	117.7 (35.9)	100.0 (0.0)	-
	Activity duration [s]	0.33 (0.18)	0.21 (0.18)	0.10 (0.02)	† (p=0.004)
M. biceps brachii left	**Mean activity [% Ref]**	459.5 (255.6)	331.7 (141.2)	60.7 (74.6)	† (p=0.004)
	Max. activity [% Ref]	853.9 (517.7)	650.8 (333.8)	100.0 (0.0)	† (p=0.002)
	Activity duration [s]	0.50 (0.32)	0.41 (0.15)	0.07 (0.09)	† (p=0.006)
M. biceps	**Mean activity [% Ref]**	335.4 (278.5)	200.5 (121.4)	41.3 (42.2)	† (p=0.014)

brachii right	**Max. activity [% Ref]**	698.8 (598.5)	383.6 (286.6)	100.0 (0.0)	
	Activity duration [s]	0.63 (0.27)	0.50 (0.15)	0.15 (0.18)	† (p<0.001)
M. triceps brachii left	**Mean activity [% Ref]**	320.0 (182.9)	912.1 (1007)	53.8 (67.6)	† (p=0.004)
	Max. activity [% Ref]	504.8 (338.1)	1747.6 (2032.4)	100.0 (0.0)	† (p=0.007)
	Activity duration [s]	0.48 (0.05)	0.54 (0.08)	0.05 (0.06)	† (p<0.001)
M. triceps brachii right	**Mean activity [% Ref]**	303.8 (90.5)	919.2 (619.0)	14.5 (43.5)	* (p=0.010) † (p<0.001)
	Max. activity [% Ref]	478.7 (148.4)	1761.1 (1324)	100.0 (0.0)	* (p=0.009) † (p<0.001)
	Activity duration [s]	0.50 (0.05)	0.55 (0.11)	0.01 (0.03)	† (p<0.001)

Figure captions

Figure 1: ToneFit Reha worn by a participant while walking

Figure 2: ToneFit Reha schematic representation

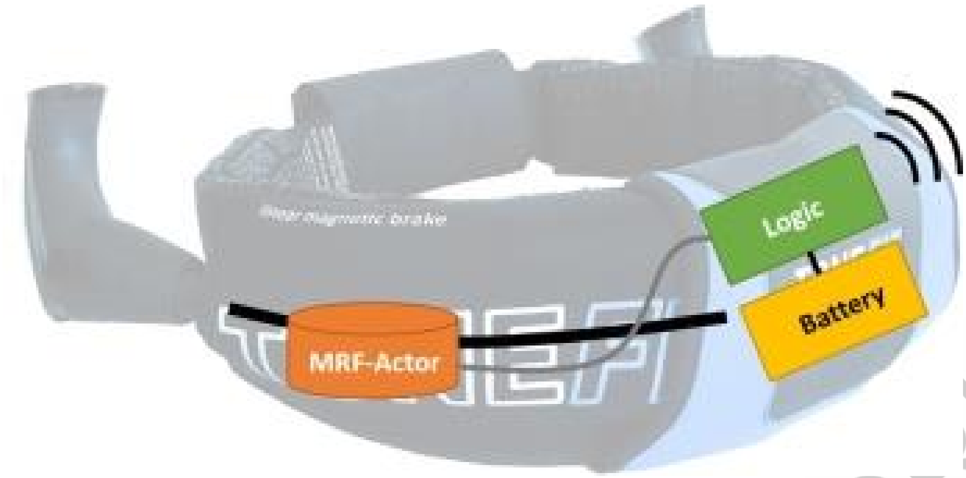


Figure 3: Trunk rotation angle to the left side [°] during the right gait-cycle; mean over 10 patients with standard deviation (shaded); vertical lines: toe off for each condition.

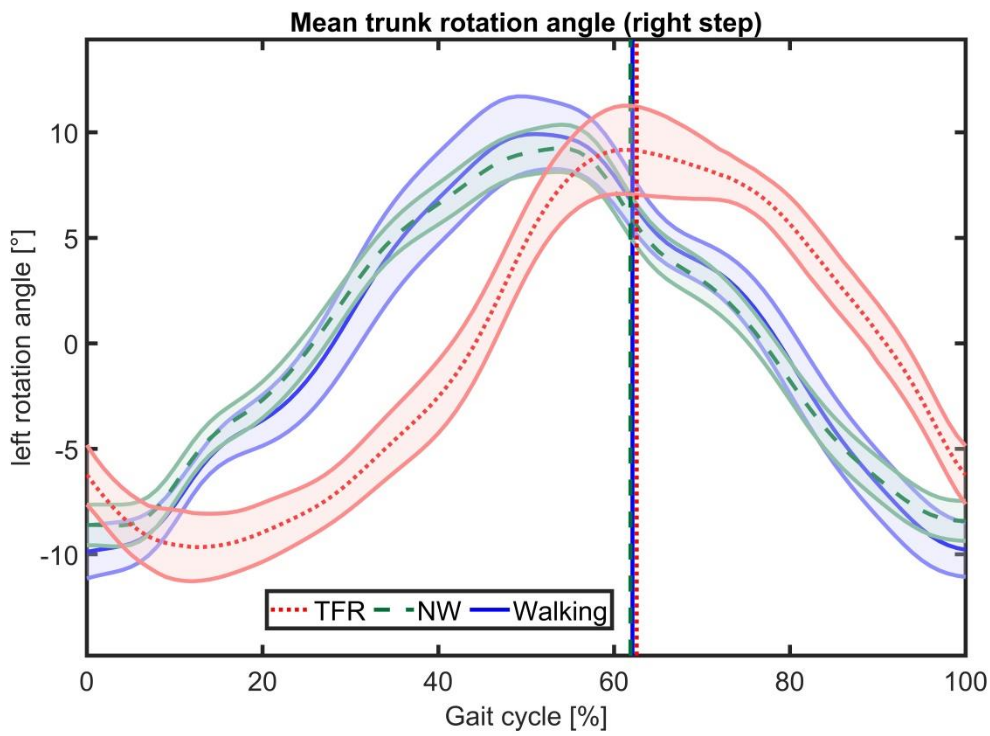


Figure 4: Right wrist ulnar deviation angle [°] during the right gait-cycle; mean over 10 patients with standard deviation (shaded); vertical line: toe off for each condition

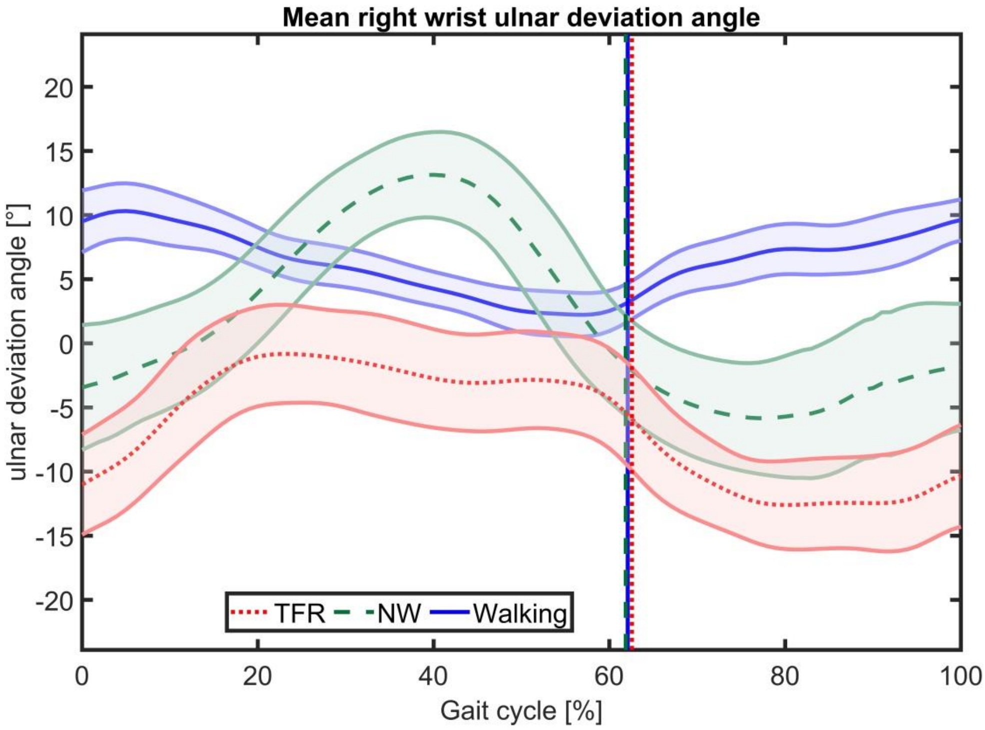


Figure 5: Right elbow flexion angle [°] during the right gait-cycle; mean over 10 patients with standard deviation (shaded); vertical line: toe off for each condition

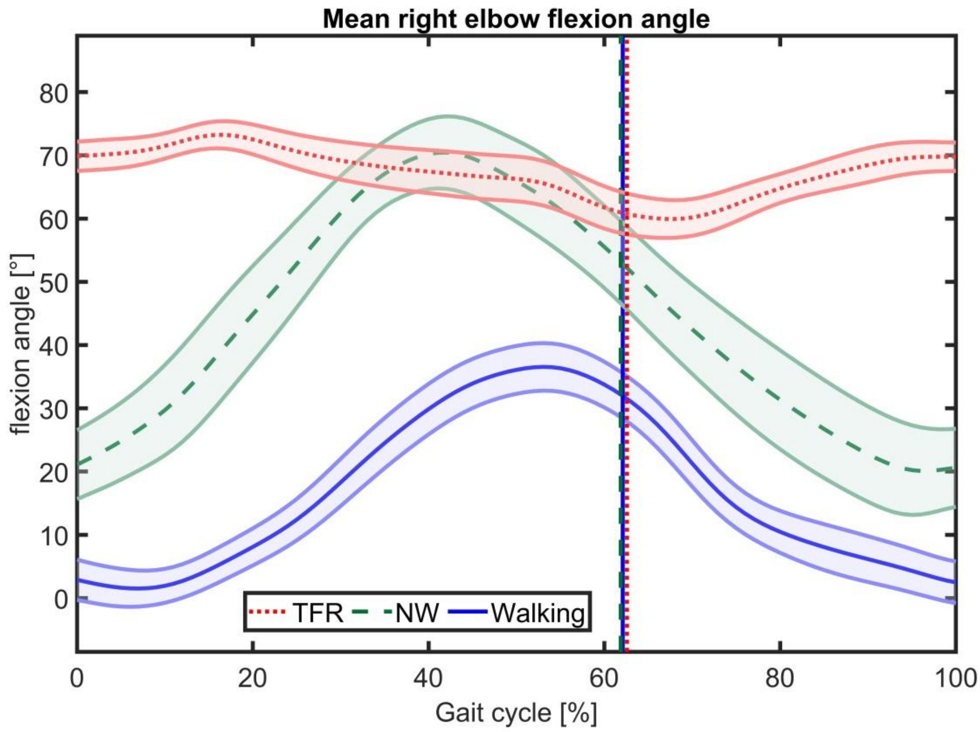


Figure 6: Right shoulder abduction angle [°] during the right gait-cycle; mean over 10 patients with standard deviation (shaded); vertical line: toe off for each condition

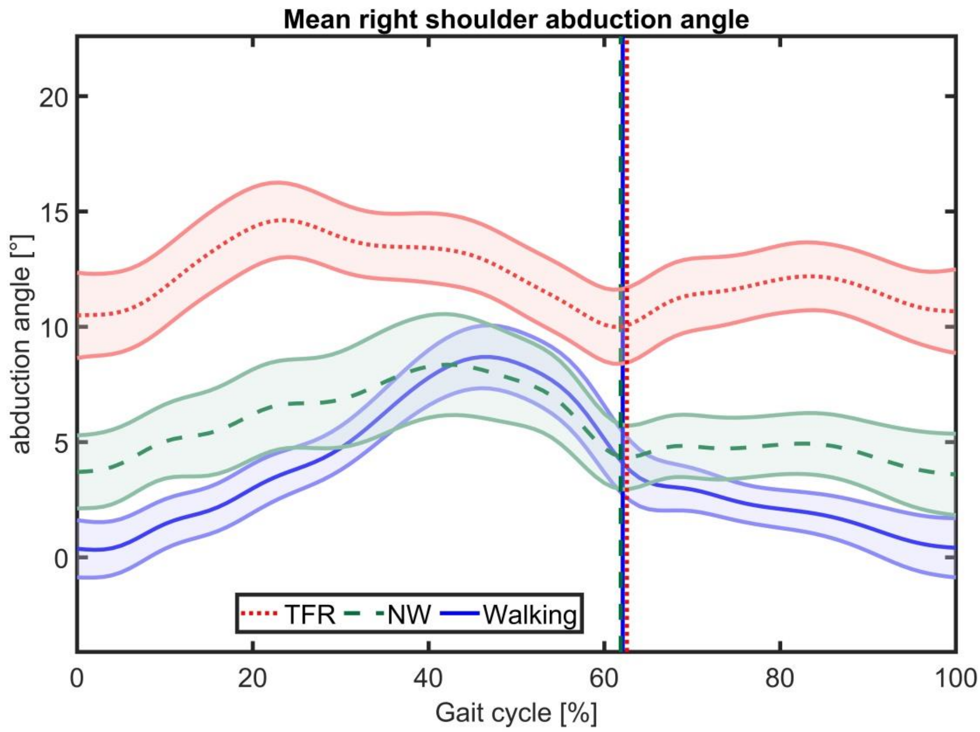


Figure 7: left ankle inversion angle [°] and left knee adduction angle [°]; mean over 10 patients with standard deviation (shaded); vertical line: toe off for each condition

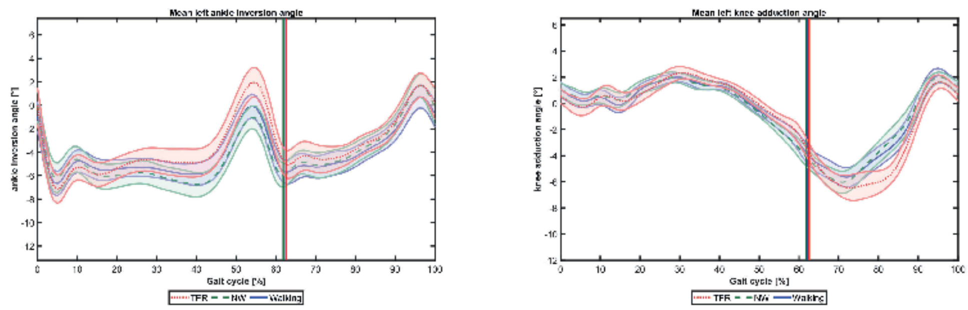
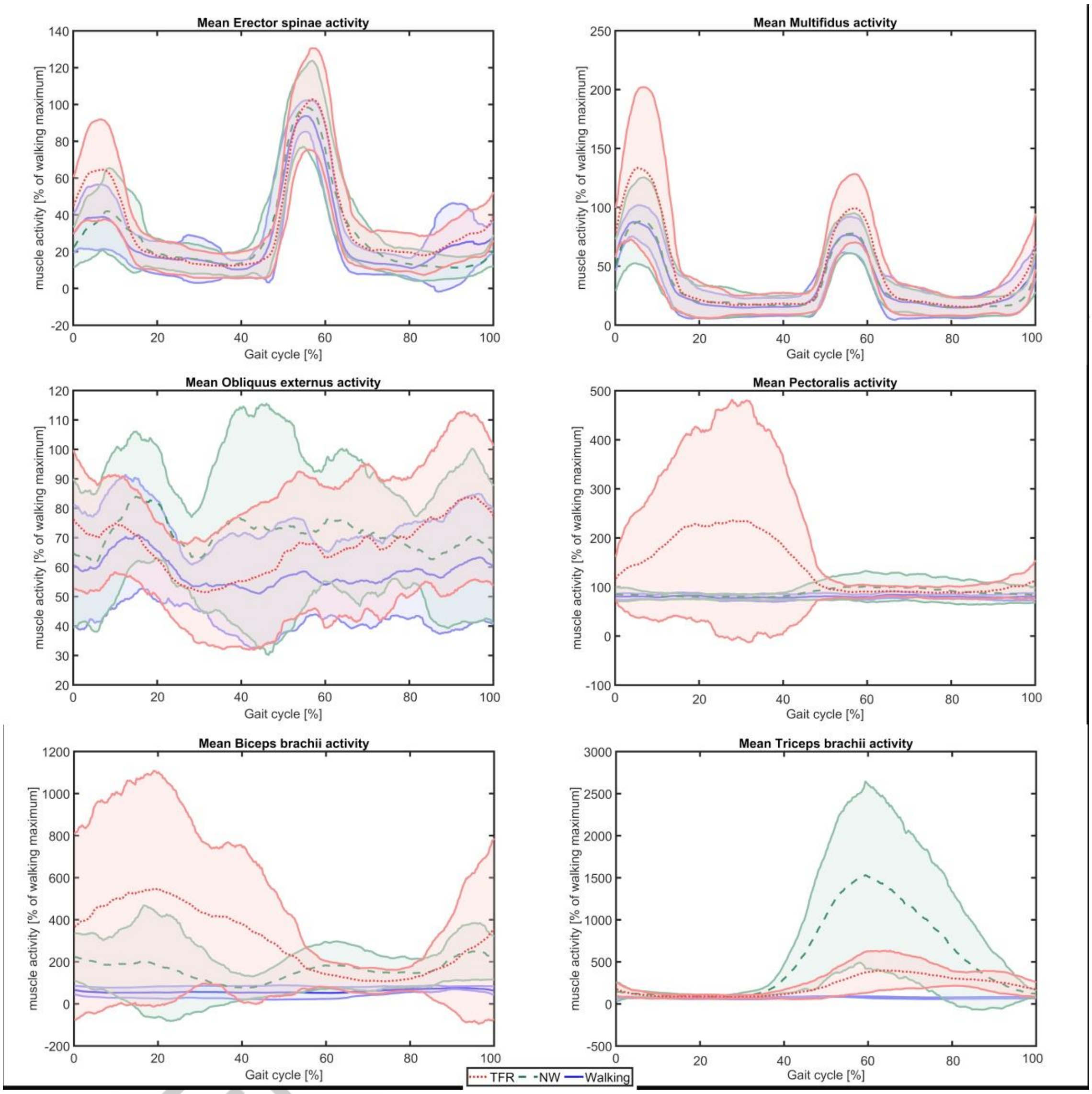


Figure 8: EMG activity for the right-side muscles. Mean of all patients, with standard deviation as shaded area.



Appendix: Marker model upper extremities

Four markers were placed non-collinearly on the upper arm and lower arm on both sides. On the dorsal aspect of both hands, three markers were placed in a non-collinear manner. The marker locations were chosen on positions with minimal skin movement. During the static trials and functional calibration trials only, additional markers were placed on the medial and lateral humeral epicondyles and the ulnar and radial styloid process. Static trials of one second duration to represent 0° of joint angle for the joints of the upper extremities (all in upright standing, feet at hip-width apart position) were recorded in the following described positions:

- Shoulder: extended elbows and wrists with fingers pointing to the floor, thumbs pointing anteriorly.
- Elbow: shoulders 90° abducted, extended elbows and wrists, thumbs pointing cranially
- Wrist: elbow in 90° flexion, extended wrists, and fingers in line with lower arm, thumbs pointing cranially

Following the static trials, functional calibration trials were recorded for each joint. For the shoulder, the joint center was calculated using the same approach as for the hip joint center [1], assuming the shoulder joint to be a ball-and-socket joint. The elbow and the wrist joints were treated as hinge joints and the functional calibrations were used to define the joint axes (calculations according to the knee joint center in List et al. [1]). Each trial included three repetitions of the described movement:

- Shoulder: circumduction with abduction and flexion of less than 90° each.
- Elbow: flexion and extension movement, starting with the elbow in 90° flexion, shoulder is in neutral position.

- Wrist: flexion and extension movement while the fingers form a fist with the thumbs pointing cranially and the elbow in 90° flexion.

The coordinate systems were defined as orthogonal, right-handed joint coordinate systems (JCS) according to Wu et al. [2]. For the shoulder joint, the JCS for the humerus relative to the thorax (section 2.4.7 of [2]) was calculated. Elbow motion was described as the motion of the forearm relative to the humerus (section 3.4.1 of [2]). For the wrist, deviations from the International Society of Biomechanics (ISB) recommendations were necessary. The axes of the wrist JCS were calculated as follows:

- e1: flexion axis, as determined in the functional calibration trial.
- e3: cross-product of e1 and a temporary axis (temp1). Temp1 is the cross-product of a temporary axis 2 (temp2) and e1. Temp2 is the line connecting the markers ElbLaTer and WriRaDia.
- e2: cross product of e3 and e1

Rotation about e1 was defined as flexion/extension; rotation about e2 as ulnar/radial deviation; and rotation about e3 as pronation/supination.

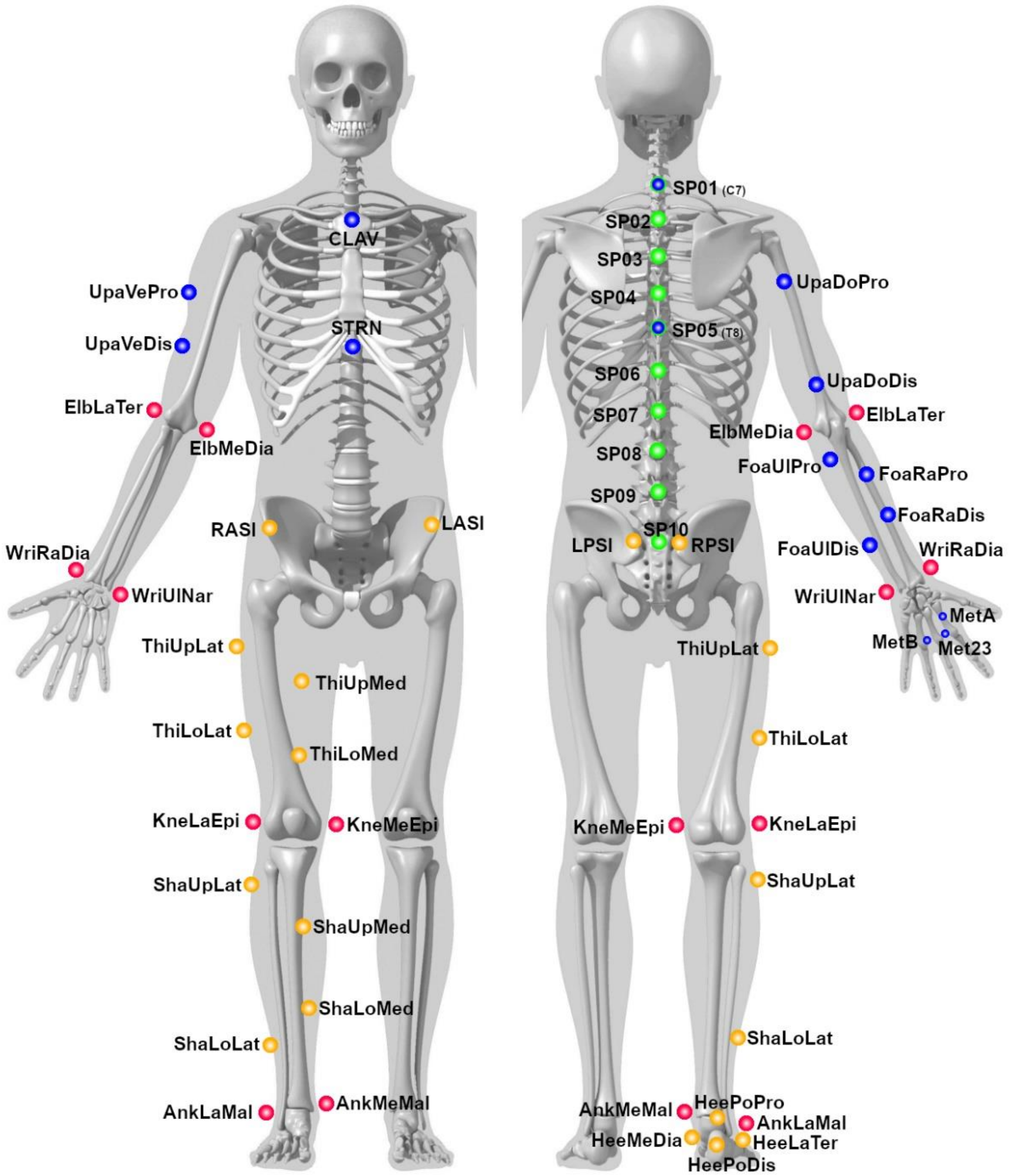


Figure A1: Marker locations of upper extremity (blue markers), lower extremity (orange markers) and trunk marker (green markers) model. Red markers indicate those that were attached only during the static and functional calibration trials.

References

[1] List R, Gülay T, Stoop M, et al. Kinematics of the Trunk and the Lower Extremities During Restricted and Unrestricted Squats: J Strength Cond Res. 2013;27:1529–1538.

[2] Wu G, van der Helm FCT, (DirkJan) Veeger HEJ, et al. ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion—Part II: shoulder, elbow, wrist and hand. J Biomech. 2005;38:981–992.