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Age-related differences in muscle activity patterns during walking in healthy individuals

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Objective: To examine how muscle activity over the entire gait cycle changes with increasing age.

Methods: Electromyography data of the erector spinae, rectus femoris, vastus lateralis, biceps femoris, tibialis anterior and gastrocnemius muscles were collected by an instrumented gait analysis during over ground walking in healthy adults aged between 20 to 89 years. Participants were categorized per decade (n=105, 15 per decade, decades 3-9). Normalized integrated linear envelopes of the electromyographic signal were calculated for one stride. A one way ANOVA using spm1d statistics explored the differences between age groups, followed by a post hoc analysis.

Results: While initiation of decline commenced at the age of 60 for erector spinae and tibialis anterior, age-related changes are most pronounced after the age of 80. Concerning timing of muscle activity, subjects in decade 7 to 9 had prolonged activity and/or early activity of the erector spinae, vastus lateralis, biceps femoris, tibialis anterior and gastrocnemius compared to other decades. Regarding amplitude of muscle activity, decreased peak amplitudes of the erector spinae, rectus femoris, vastus lateralis and gastrocnemius were observed in decades 7 to 9 compared to other decades.

Conclusion: Both timing and amplitude of muscle activation patterns need to be considered to understand the aging process. Regarding the erector spinae, tibialis anterior and vastus lateralis, a decrease in muscle activation coincides with prolonged activity, compared to the gastrocnemius where decreased muscle activation is associated with early activation.

Introduction

Aging alters muscular behaviour which results in the use of different muscle strategies during walking compared to younger adults. Older subjects rely more on proximal, rather than distal muscles, for power generation [1-3]. A redistribution of the relative contribution of individual muscle groups to the total output was seen in older individuals compared to younger adults. Additionally, an increase in muscle activity was found in the elderly during quiet standing and walking which suggests that high levels of muscle activity are a characteristic of age-related declines [4-6]. For example, Lee et al (2017) reported that elderly adults (76 ± 3 years) showed significantly higher lower limb muscle co-activation than the young (24 ± 2 years), and middle-aged adults (53 ± 2 years) during gait at a self-selected speed [6]. The elderly showed increased activation of the tibialis anterior and soleus during mid stance, and greater activation of the vastus lateralis and medial hamstring during loading response and mid-stance [3]. Moreover, during stance phase increased co-activation was mostly seen in the thigh muscles compared to swing phase where greater levels of co-activation were present in the shank [6]. During stance phase the leg is being loaded and maintaining stability surrounding the knee is of great importance which can be related to increased activity in the thigh. In contrast to the swing phase were the limb is not loaded but were push off of the plantar flexor muscles is important to achieve propulsion in combination with preposition of the foot in late swing.

The current available research uses a cut off of at the age of 60 years to categorize individuals into two age groups, young and older adults. Yet, changes in gait pattern still occur after this age [7,8]. In 25 percent of individuals between 70-74 years of age, gait changes were detected, compared to nearly 60 percent of those between the ages 80 to 84 years [8]. Therefore, age groups should be divided into smaller samples, so that age-related changes can be located more precisely. Instrumented three-dimensional motion analysis that provides quantitative measures is the gold standard for gait assessment. Gait analysis can be executed on a treadmill or over ground, at a self-selected or predetermined walking speed. Since self-selected walking speed is an indicator for health problems and can serve as a predictor for future health status [9,10], it is important to incorporate this variable during the instrumented gait analysis.

Investigating age-related changes in more than two age groups, might not only help clinicians to redirect therapy goals, it can also help in determining appropriate intervention strategies in older individuals to counteract early decline and to reduce risk of falls. Since muscle activation patterns, such as prolonged activity and increased co-contractions, are able to distinguish between fallers and non-fallers [11]. Several authors have suggested that this increase in muscle activity is a compensatory strategy to enhance stability and decrease the fear of falling [12].

Thus, the purpose of this present study was to examine age-related changes in muscle activity across the entire gait cycle during over ground walking at a self-selected speed assessed by instrumented three-dimensional gait analysis. Furthermore, the study sample will be divided into multiple small age groups to allow identification of the onset of deterioration regarding muscle activity.

Methods

Setting

Participants received an instrumented gait analysis performed at a movement analysis laboratory equipped with three-dimensional motion capture system with eight cameras (Vicon T10, 100 Hz., ©Vicon Motion Systems Ltd., Oxford, UK, 100 fps, resolution 1 Megapixel (1120 x 896), 3 AMTI type OR 6-7 force plates (1000 fps, 46x50x8 cm) and 1 AccuGait® (1000 fps) force plate. Reflective markers were attached to anatomical landmarks on the participant's body according to the standard Plug-In-Gait model [15]. Muscle activity was recorded with 16 channel telemetric wireless surface electromyography (EMG) system (Aurion Zerowire®, Cometa, Rome, Italy, 1080 Hz) in combination with 3M™ Red Dot Monitoring Electrodes (circular electrodes, 60 mm diameter).

Participants walked barefoot over a 12 meter walkway at a self-selected speed (0.52-1.55 m/s). In total, a minimum of six valid walking trials were recorded, indicating that visibility of the markers and clean heel strikes of the left and right stride were present. The study protocol was approved by the local ethics committee (B300201316328). The period of data collection was between April 2015 and January 2016.

Participants

Subjects participated on a voluntary basis. Eligibility criteria for participation were adults between 20 to 89 years, categorized into seven decades. Participants were excluded if they had self-reported visual impairments, antalgic gait pattern, abnormal mobility in the lower limbs or any known neurological or orthopaedic disorder that could influence motor performance and balance based on questionnaires. Informed consent was obtained from all subjects prior to participation.

Measurements and data calculations

For each subject, body mass, height, leg length (from spina iliaca anterior superior to medial malleoli) and joint width (ankle, knee, elbow, wrist and finger) were collected according to standard procedures [13]. Participants were properly prepared before attaching the electrodes by shaving and cleaning the skin to ensure a good electrode-skin contact. EMG electrodes were placed according to the SENIAM guidelines, oriented parallel to the muscle fibers with an inter-electrode distance of 20 mm [14]. Reflective markers were tracked and labelled using the Vicon Nexus 1.8.5 software. Based on the ankle marker trajectories and force plate recordings, events of foot strike and foot off were determined. The gait cycle was calculated based on the left and right heel marker trajectories. Spatiotemporal parameters (STP) were calculated with the 'Generate Gait Cycle Parameters Pipeline Operation' of the Vicon Nexus software. When EMG data was reliable for at least three consecutive strides, trials were further processed. This meant that the raw EMG data was larger than the noise of the signal (good signal-to-noise ratio), that all markers were visible and that a clean heel strike and toe off were recorded. The raw data (c3d files) were exported to a custom made Matlab (R2015a for Windows) model to calculate the variables of interest. Due to the high amount of data, only muscle activity and STP of the right side during a right stride was further analysed.

Variables of interest

The variables of interest consisted of STP and the normalized integrated linear envelope of the EMG signal of the m. erector spinae, m. rectus femoris, m. vastus lateralis, m. biceps femoris, m. tibialis anterior, the lateral head of the m. gastrocnemius. A variety of STP were investigated: step time (s), step length (m), step width (m), stance (%) and walking speed (m/s). Step time is the time between contralateral and ipsilateral foot contact. Step length is the linear distance between contralateral and ipsilateral foot contact. Step width is the side-to-side distance between the midpoint of the heel of both feet. Stance is the amount of the time foot is in contact with the ground, it is calculated as the percentage of the gait cycle at contralateral foot off. At last, walking speed is calculated as stride length divided by stride time. The EMG signal was computed based on raw EMG data and was full-wave rectified to obtain absolute values of the signal. Thereafter the linear envelop, integrated EMG and normalized EMG was acquired by computing the outline of the signal, the area under the curve and calculating the average EMG signal throughout the gait cycle (1000 points per stride) for every subject respectively. Amplifier gain of the EMG signal was set at 1 Volt. EMG signals were band-pass filtered (10–300 Hz), rectified, smoothed using a 50 msec moving average window to generate a linear envelope, and normalized to 1000 points per stride. To create a bandpass filter, the EMG data were first filtered with a 2nd order zero-phase butterworth high-pass filter with a cut-off frequency of 10 Hz. Then a 2nd order zero-phase butterworth low-pass filter was used with a cut-off frequency of 300 Hz. The linear envelope was calculated using the average of a 50 msec moving window and was normalised to mean amplitude over the gait cycle. Graphs were plotted to visualize the EMG activity (x = 1000 points per stride/gait cycle; y = EMG activity normalized to mean), a y-value of one represents mean muscle activity calculated over the entire gait cycle. This method has also been reported by Schmitz et al (2009) [3]. Both timing and amplitude of the aforementioned muscles were observed. A difference in timing was defined as EMG activity greater than 1 compared to less than 1 in other decades. Amplitude of the EMG signal was assessed during activity bursts (peak amplitude). The gait cycle was divided into six phases: loading response (0-10% of gait cycle), mid stance (10-30% of gait cycle), terminal stance (30-50% of gait cycle), initial swing (60-70% of gait cycle), mid swing (70-85% of gait cycle), and terminal swing (85-100% of gait cycle).

Statistical analysis

Statistical analysis of the subject characteristics and STP were examined by means of one way ANOVA, Tukey HSD examined post hoc differences, and was performed with SPSS version 23® for Windows (©IBM Corporations, New York, USA). Muscle activity have previously been examined using traditional zero-dimensional statistics by selecting parameters at several key events instead of looking at the entire gait cycle. This type of statistics examines EMG patterns, which are continuous data sets during walking, as a mean/maximum/minimum amplitude, mean EMG pattern or as a coefficient of variation. As a result, a great amount of data points are being created which increases the risk of false positive results. Zero-dimensional statistical significance in kinematic trajectories is easily achieved when there is, in fact, no one-dimensional effect. On the other hand, one-dimensional statistics incorporates time series and examines the entire trajectories. Adopting this fairly new technique, one-dimensional Statistical Parametric Mapping (spm1d), can more tightly control for false positive results [15,16]. This technique not only detects changes in continuous data sets, it also conducts statistical hypothesis testing in a continuous manner, directly on the original curves. By doing this, spm1d is able to overcome the aforementioned limitations of zero-dimension statistics and has an added value by investigating both timing and amplitude simultaneously. Spm1D has been thoroughly described by Pataky et al [15,16]. Normalized integrated linear envelope of the EMG signal were assessed across the entire gait cycle between decades by means of spm1d, one way ANOVA. If the null hypothesis is true, the same variances and smoothness would produce identical curves. The null hypothesis was rejected when the F value/*t* value exceeded the critical test statistical value α. Post hoc analysis consisted of two-sample t-tests conducted on all group pairs. The F statistics compares the joint effect of all the variables together, in contrast to the t-statistics which confirms where the differences between groups occurred. Significance level was set at $p<0.05$.

Results

Participants and descriptive data

In total, 105 participants were included in this study, 15 per decade. Subject characteristics can be found in Table 1. No significant differences were found in sex distribution, leg length and body mass ($p>0.05$) but subjects from decade 9 were significantly shorter than subjects in decades 3 and 4 ($p=0.044$ and $p=0.002$, respectively).

Table 1. Subject Characteristics

[Insert table 1]

Spatiotemporal parameters

Significant differences were found in walking speed and step length ($p<0.001$), subjects in decades 8 and 9 walked slower compared to the other decades ($p<0.05$). Step length decreased throughout all decades, participants in decades 7 to 9 walked with significant smaller steps compared to the other decades ($p<0.05$). No significant overall differences were found between stance, step time and step width. An overview of the STP can be found in Table 2.

Table 2. Spatiotemporal parameters

[Insert table 2]

Muscle activity

The normalized integrated linear envelope of the EMG signal and the statistical analysis of the EMG patterns using spm1d technique can be found in Figure 2.

Figure 1. Comparison of muscle activity between decades

[Insert figure 1]

M. Erector spinae

Gait Pattern. All decades showed a similar biphasic pattern with a peak at ipsilateral and contralateral foot contact. Significant differences between decades were observed at initial contact ($F=4.019$, $p=0.043$), during mid-stance, terminal stance and pre-swing ($F=4.019$, $p<0.001$).

EMG Timing. Participants in decade 7 to 9 showed prolonged activity of m. erector spinae after heel strike compared to decades 3 to 6. The ES of the subjects in decade 9 remained active the longest compared to subjects in decade 7 ($t=3.493$, $p<0.001$) and 8 ($t=3.509$, $p<0.001$).

EMG Amplitude. The amplitude at ipsilateral foot contact, only subjects in decade 9 showed decreased activity compared to other decades in contrast to contralateral foot contact where participants in decades 7 to 9 had lower activity levels, no difference were seen between the three last decades at contralateral foot contact.

M. Rectus femoris

Gait Pattern. All decades showed a similar biphasic pattern with a large peak during loading response and a small increase from pre-swing until initial swing. Significant differences between decades were observed at opposite toe off (F=4.003, p=0.005), during the last part of terminal stance, pre-swing and mid swing (F=4.003, p<0.001).

EMG Timing. During the first peak at initial contact, no differences were observed. However, the second small increase, circa terminal stance and pre swing, occurred sooner and was prolonged in participants in decades 5 and 7 compared to the other decades.

EMG Amplitude. There was no rise in m. rectus femoris activity in subjects in decade 8 and 9 during pre-swing and initial swing and they had significantly lower amplitudes than decades 3 to 7. At opposite toe off, subjects in decade 8 had a significantly higher amplitude than subjects in decade 4 (t=3.490, p=0.024), 5 (t=3.497, p=0.042), and 7 (t=3.526, p<0.001).

M. Vastus lateralis

Gait Pattern. All decades showed a similar monophasic pattern with a maximal peak during loading response. Significant differences were found between decades during mid stance (F=3.955, p<0.001), the last part of terminal stance (F=3.955, p<0.001), pre swing (F=3.955, p<0.001), a small part of initial swing (F=3.955, p=0.011) and terminal swing (F=3.955, p=0.033).

EMG Timing. Participants in decade 8 and 9 showed prolonged activity of the m. vastus lateralis during mid stance compared to other decades.

EMG Amplitude. Only significant differences were observed during pre-swing and initial swing, subjects in decades 8 and 9 had decreased activity of the m. vastus lateralis compared to decade 3 to 5. Yet, muscle amplitude in participants in decade five also significantly differed during this phase from decade 6 (t=3.440, p=0.001).

M. Biceps femoris

Gait Pattern. All decades showed the similar monophasic pattern with a maximal peak at initial contact. Significant differences in EMG activity between decades was seen during mid stance (F=3.990, p<0.001), during pre-swing (F=3.990, p=0.038), initial swing (F=3.990, p=0.025) and terminal swing (F=3.990, p<0.001).

EMG T*iming.* Participants in decade 9 showed prolonged activity of the m. biceps femoris after initial contact compared to decades 3 (t=3.504, p=0.003), 4 (t=3.472, p=0.050) and 5 (t=3.433, p<0.001). At last, post hoc tests revealed that decade 3 and 9 showed delayed activity of the m. biceps femoris during terminal swing compared to decades 4, 6 and 7.

EMG Amplitude. A small burst of activity was observed during pre-swing in subjects in decade 3 which resulted in significant differences compared to decades 7 (t=3.510, p=0.049) and 8 (t=3.525, p=0.007). The observed differences during initial swing was due to a rise of activity in the subjects in decade 5 compared to subjects in decade 4 (t=3.470, p<0.001) and 9 (t=3.433, p=0.009).

M. Tibialis Anterior

Gait Pattern. All decades showed the similar biphasic pattern with a peak at initial contact and initial swing. Significant differences were observed during loading response (F=3.990, p=0.037), mid stance (F=3.990, p<0.001), terminal stance (F=3.990, p<0.001), mid swing and terminal swing (F=3.990, p<0.001).

EMG Timing. Subjects in decades 7 to 9 showed prolonged activity of the m. tibialis anterior after initial contact compared to decades 3 and 4.

EMG Amplitude. During terminal stance, subjects in decade 8 had a significantly lower amplitude compared to all other decades. Additionally, lower amplitudes of the m. tibialis anterior were also observed in participants from decade 9 compared to decade 3 (t=3.491, p<0.001) and 4 (t=3.503, p<0.001). During mid swing subjects in decade 7 to 9 showed lower amplitudes of m. tibialis anterior compared to decades 3, 4 and 6.

M. Gastrocnemius

Gait Pattern. All decades showed the similar monophasic pattern with a peak during the transition of terminal stance to pre-swing. Significant differences were present at initial contact (F=3.926, p=0.046), mid stance (F=3.926, p<0.001), terminal stance (F=3.926, p=0.003).

EMG Timing. Early activation of the m. gastrocnemius was seen in decades 8 and 9 compared to the other decades. In addition, subjects in decade 4 for also showed premature activity compared to decades 6 (t=3.444, p=0.002) and 7 (t=3.440, p=0.015).

EMG Amplitude. During push off, participants in decade 3 had a significantly larger amplitude compared to the other decades in contrast to the subjects in decade 9 who had significantly smaller amplitudes compared to the others.

Discussion

The purpose of the present study was to examine age-related changes in muscle activation patterns across the entire gait cycle using spm1d and by dividing the study sample into multiple small age groups to allow identification of the onset of deterioration and comparison of the continuous dataset regarding muscle activity. Our findings suggest that prolonged and early activation in combination with decreased muscle amplitude was already present in decade 7 (60-69 years), but was most prominent in decade 9 (80-89 years) during walking at a self-selected speed. More specifically, subjects in decade 7 to 9 had prolonged activity and/or early activation of all muscles except for m. rectus femoris, in combination with decreased peak amplitudes at key events in the gait cycle of all the muscles except for m. vastus lateralis.

Several authors have suggested that the amplitude of the EMG signal increases with age during quiet standing [4], during treadmill walking at several controlled speeds [5] and during over ground walking at a self-selected speed [3] measured with two or three age groups, which seems in contrast to our findings. Since we examined our subjects during over ground walking at a self-selected speed, we compared our results to those of Schmitz et al (2009) [3]. They assessed age-related changes in two age groups on the activation of lower extremity muscles during over ground walking on a 12 meter walkway at a self-selected speed [3]. Although the design of the study was similar, they computed the average normalized EMG activity within selected phase of the gait cycle, in contrast to our study where we compared over the entire gait cycle. However, by averaging, no distinction can be made between timing and amplitude of muscle contractions. For example, during mid stance, increased activity of the m. vastus lateralis was reported in the elderly (73 ± 5 years) [3]. Looking more closely at our curves comparing the several decades (Figure 1), the increase in activity during mid stance is actually originating from prolonged activity of the m. vastus lateralis after its activity burst during loading response. So, previous studies compared EMG amplitude between young and old individuals, regardless of the timing of muscle contractions. Since on and off times are mostly used in clinical practice to assess the EMG signal, emphasizing the importance of timing of muscle contractions compared to amplitude during EMG measurement seems crucial.

The most important differences in timing occurred during loading response, prolonged muscle activity was seen after this phase in decade 7 (60-69 years) for the m. erector spine and m. tibialis anterior, in decade 8 (70-79 years) for the m. vastus lateralis, and in decade 9 (80-89 years) for the m. biceps femoris. During loading response weight acceptance of the ipsilateral limb takes place resulting in an inherent state of instability. This instability is counteracted by a burst of muscle activity to ensure sufficient stability during walking, which requires both cognitive and sensorimotor processes [17]. However, aging has been associated with a progressive decline of motor behavior, perception and cognition [18,19]. It might be that elderly subjects compensate for this decline by prolonging their restabilization time. Another possible explanation might be the interaction between timing and amplitude of muscle activity, that forms an integrated whole. Concerning muscle amplitude, our most important finding is that decreased muscle activity was seen during activity bursts from the age of 60 years in the erector spinae and m. tibialis anterior, from the age of 70 years in m. rectus femoris and m. vastus lateralis, and from the age of 80 years in the m. gastrocnemius. Regarding the m. erector spinae, m. tibialis anterior and m. vastus lateralis a decrease in muscle activation coincides with prolonged activity, compared to the m. gastrocnemius where decreased muscle activity is associated with early activation. A change in muscle timing could therefore be a compensatory strategy for the decrease in muscle amplitude during over ground walking. The relationship between timing and amplitude was not that clear in m. rectus femoris and m. biceps femoris as significant changes in amplitude were seen, their biological relevance is questionable given the small amplitude (< 1) across all decades. No noticeable differences in timing across decades during over ground walking at a self-selected speed were observed in the m. rectus femoris. On the other hand, the m. biceps femoris does show important differences in timing, e.g. prolonged activity after loading response and early activation in terminal swing, without clear changes in muscular amplitude.

As already mentioned, differences in timing, e.g. prolonged activity, are primarily seen during phases of instability when walking over ground. Although co-activation, as suggested by several authors, seems of great importance to explain the differences in muscle activation patterns [4,20,21], the neurophysiological involvement should not be neglected. Since the on and off timings of muscle activations (contraction-relaxation cycles) of the respective muscles are changed in the elderly, muscles which are generally not activated simultaneously, e.g. agonist and antagonist muscles, are now activated at the same time. For example, due to prolonged activity of the m. tibialis anterior and early activation of the m. gastrocnemius during stance in the elderly, both muscles are activated simultaneously during mid stance when walking over ground at a self-selected speed (Figure 1). This could be a strategy to increase joint stiffness and enhance stability to compensate for several neuro-motor impairments associated with aging [22]. However, excessive co-activation in elderly adults is likely to increase the energy cost of locomotion, thereby inducing fatigue and increasing the risk of falling [23].

To understand the changes in amplitude, one could look at the cellular level of the muscle. Aging is associated with a loss of muscle mass, defined as sarcopenia, in combination with a cellular atrophy in type II, fast twitch, muscle fibers [24]. Between the ages of 24 and 80 years a total diminution of 40% in muscle size is apparent which translates in a 0.5-1% decrease in whole muscle cross-sectional area per year beyond the fifth decade [24]. Moreover, the type II fibres experience a 26% loss in cross-sectional area when subjects in the third and ninth decades where compared to each other, while type I fibres did not differ between these individuals [24]. Due to a decrease in type II fibres, muscles tend to contract slower and generate less force which could translate in lower peak amplitudes, and even in early activity as a compensation for slower contraction times. This might explain why in the gastrocnemius early activation coincides with decreased amplitude during push off when walking over ground at a self-selected speed. Although early activation can be explained by a change in fibre type, prolonged activity cannot since the type of fibre predominating in muscle tissue does not correlate with relaxation times [25]. On the other hand, aging also alters the biomechanical response of the muscle-tendon unit. Many cellular changes occur in an aging tendon, density and number of tendon cells per unit of surface decreases, tendon cells become longer and thinner and the metabolic activity of tenoblasts decrease [26]. Those tendon changes to correlate with changed relaxation times, according to Plate et al. (2013) decreased relaxation times were seen in aged tendons compared to younger tendons in a rat model [27].

The decrease in muscular activity was most apparent from the age of 80 years which might be related to previously observed changes in joint kinematics [28]. Observation of changes in gait kinematics and kinetics as a result of age were specific for the anatomical plane under investigation: age-related changes in the frontal plane were apparent from the decade 7 (60-69 years), in the transverse plane from the decade 8 (70-79 years) and in the sagittal plane from the decade 9 (80-89 years) [28]. Subjects in decade 9 (80-89 years) walked with a more flexed gait pattern resulting in increased internal and external moments. The decrease in amplitude of the anti-gravity muscles could explain why the elderly are less able to maintain an upright position during walking over ground than younger individuals. Although a direct quantification of muscle strength by means of EMG is not appropriate, in most cases, the EMG signal increases non-linearly with increasing force [29]. It is therefore a plausible hypothesis that muscle weakness, as a result of aging, has a direct effect on age-related changes in gait kinematics. However, it is difficult to draw conclusions for the muscle responsible for movements in the frontal and transversal plane, since they were not investigated.

Although this study adds to the current knowledge of age-related changes in EMG patterns in adults by giving a more comprehensive representation of the observed changes, there are some important limitations to consider. The first limitation is the influence of gait speed, a higher walking speed is related to an increase in ankle plantar flexion and knee/hip flexion which could influence EMG activity during over ground walking [30]. Some observed differences might be the result of variations in walking speed between decades concerning peak amplitudes, yet no differences should be expected concerning timing of muscle activity since no changes in key gait events were apparent. For future research, it is of interest to examine whether the differences over the entire gait cycle are still present when walking at an equivalent gait speed. In addition, although abdominal muscles play a major role when talking about stability, we did not investigate them. Previous research experience in our laboratory revealed that that EMG data derived from abdominal muscles are not reliable. The presence of subcutaneous fat not only reduces EMG amplitude, but also increases cross-talk which increases the noise of the signal [31,32]. The risk of subcutaneous fat around the abdominal muscles is rather large. Subsequently, since smp1d is a fairly new statistical manner to examine a continuous dataset, post hoc analyses are still under development and should be interpreted with caution. Yet, resulting errors are expected to be small.

Clinical implications

While initiation of decline commenced at the age of 60 years for m. erector spinae and m. tibialis anterior, age-related changes are most pronounced after the age of 80 years. While these results are partly similar to other studies concerning aging [3-5,33], age-related changes present themselves later than generally expected. Although some changes are rather small, the most prominent changes were seen in the m. erector spinae which remained more active throughout the entire stance phase in the elderly during over ground walking at a self-selected speed.

There is indeed a distal to proximal shift in muscle activity. We believe that training both the proximal and distal muscles groups are of priority. Since the elderly rely more on proximal muscles groups, it is important to increase proximal control to reduce fall risk by strengthening trunk musculature. However, it is also of great importance to counteract early decline and generating a sufficient amount of power to enhance propulsion which is mainly generated by the distal muscles such as m. gastrocnemius. Several studies have documented that strength training can increase EMG amplitude of agonist muscles and decrease co-activation of antagonist muscles [34]. Moreover, strength training is able to improve onset-rates of EMG which represents an increase in rapid motor units firing, earlier recruitment of motor units and augmented motor unit synchronization [34]. Moreover, excessive co-activation in elderly adults is likely to increase the energy cost of locomotion, thereby inducing fatigue and increasing the risk of falling. Integrating specific muscle strengthening programs starting before the age of 60 years and continuing afterwards might help us in counteracting the decline and decrease fall risk.

Conclusion

Results showed age-related changes in muscle activity, subjects in decades 7 to 9 showed prolonged and early activity in combination with decreased peak amplitude in the majority of muscles during over ground walking at a self-selected speed, which emphasizes the importance of investigating both timing and amplitude of muscle contractions during EMG measurements. Both timing and amplitude of muscle activation patterns seem to form an integrated whole in understanding the aging process. Although a clear decline in muscle activation was only seen from the age of 80 years, subjects in their 60’s and 70’s already showed a tendency towards decline which was most prominent in the trunk muscles.

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Conflict of interest:

The authors have no conflicts of interest to declare.

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Figures and Tables

Table 1. Subject Characteristics

Table 2. Spatiotemporal parameters

Figure 1. Comparison of muscle activity between decades

Table 1. Subject Characteristics

Decade (n=105, 15 per decade)	Age (y)	Gender (F/M)	Body Height (mm)	Body Mass (kg)	Leg Length (mm)
3 (20-29y)	25 ± 2	9/6	1735 ± 90*	73 ± 14	923 ± 50
4 (30-39y)	33 ± 2	7/8	1757 ± 95*	84 ± 32	935 ± 66
5 (40-49y)	45 ± 2	8/7	1674 ± 118	74 ± 15	890 ± 79
6 (50-59y)	53 ± 2	8/7	1690 ± 85	68 ± 14	891 ± 46
7 (60-69y)	63 ± 2	9/6	1660 ± 78	77 ± 11	900 ± 44
8 (70-79y)	73 ± 2	8/7	1653 ± 111	75 ± 9	890 ± 87
9 (80-89y)	82 ± 1	6/9	1618. ± 115	74 ± 11	890 ± 69

***Tukey HSD, Significant differences with decade 9, $p<0.05$**

y: years; F: female; M: Male; n: number of subjects; mm: millimetres; kg: kilograms;

Table 2. Spatiotemporal parameters

Decade	Stance (%)	Walking Speed (m/s)	Step time (s)	Step length (m)	Step width (m)
3 (20-29y)	60.41 ± 1.48	1.19 ± 0.14[b]	0.55 ± 0.04	0.66 ± 0.06[c]	0.17 ± 0.02
4 (30-39y)	59.53 ± 1.09	1.25 ± 0.10[b]	0.53 ± 0.08	0.66 ± 0.05[c]	0.19 ± 0.05
5 (40-49y)	59.82 ± 1.66	1.21 ± 0.11[b]	0.51 ± 0.04	0.64 ± 0.06[c]	0.18 ± 0.02
6 (50-59y)	60.06 ± 1.85	1.21 ± 0.19[b]	0.52 ± 0.04	0.65 ± 0.07 [c]	0.17 ± 0.03
7 (60-69y)	60.55 ± 2.33	1.22 ± 0.13[b]	0.52 ± 0.03	0.62 ± 0.06	0,17 ± 0.02
8 (70-79y)	60.24 ± 5.30	1.07 ± 0.11	0.53 ± 0.03	0.59 ± 0.09	0.17 ± 0.04
9 (80-89y)	61.37 ± 1.47	1.04 ± 0.19	0.54 ± 0.05	0.56 ± 0.09	0.19 ± 0.03
Total	60.28 ± 2.53	1.17 ± 0.18[a]	0.53 ± 0.04	0.63 ± 0.77[a]	0.18 ± 0.03

[a] ANOVA, $p \leq 0.001$

[b] Tukey HSD, differences with decade 8-9, $p<0.05$

[c] Tukey HSD, differences with decade 7-8-9, $p<0.05$

y: years; %: percentage; m: metres; s: seconds

Figure 1. Comparison of muscle activity between decades

(see additional file)