Gluteus medius: an intramuscular EMG investigation of anterior, middle and posterior segments during gait

1 Introduction

The gluteus medius (GMed) muscle is considered the prime abductor of the hip joint (Standring et al., 2005), with its main function in weight bearing to stabilise the pelvis in unilateral stance against the effects of gravity (Al-Hayani, 2009; Gottschalk et al., 1989). Cadaveric studies suggest that gluteus medius (GMed) is comprised of three structurally unique regions (anterior, middle and posterior) (Al-Hayani, 2009; Gottschalk et al., 1989; Semciw et al., 2012a) with potential for independent control from the central nervous system (CNS) (Gottschalk et al., 1989; Soderberg and Dostal, 1978). This has led researchers to consider a broader role of GMed, by attributing a role of pelvic rotation, in addition to pelvic stability for anterior and posterior GMed (Al-Hayani, 2009; Gottschalk et al., 1989).

There are a number of studies that have attempted to assess the function of three segments of GMed with electromyography (EMG) (Gottschalk et al., 1989; O'Dwyer et al., 2011; O'Sullivan et al., 2010; Soderberg and Dostal, 1978). By using surface electrodes, three studies concluded that each segment has the capacity for independent activity in isometric tasks (O'Dwyer et al., 2011), weight-bearing exercises (O'Sullivan et al., 2010) or gait (Gottschalk et al., 1989). Unfortunately a number of methodological limitations bring into question these conclusions. First, the use of surface electrodes for all three segments (Gottschalk et al., 1989; O'Dwyer et al., 2011; O'Sullivan et al., 2010) is inappropriate since posterior GMed is completely covered by gluteus maximus (GMax) (Hodges et al.,

1997; Semciw et al., 2012a). Furthermore, myoelectric activity recorded from surface electrodes may be contaminated by cross talk from surrounding muscles (Bogey et al., 2000; Chapman et al., 2006; 2010; Johnson et al., 2011; Perry et al., 1981), therefore activity from middle or anterior segments may be contaminated given their proximity to the surrounding GMax and tensor fascia latae (TFL) muscles (Semciw et al., 2012a). The investigators of a fourth study partitioned GMed into three segments, and inserted intramuscular electrodes into these regions without real time ultrasound (RTUS) guidance (Soderberg and Dostal, 1978). Although the authors reported phasic activity of GMed in a range of functional tasks, verified guidelines for unique segments of GMed were not used. It is therefore unclear as to whether electrodes were accurately inserted into functionally unique segments of GMed or possibly other muscles.

Two recent reviews report on GMed function, as determined by EMG, in a range of commonly prescribed rehabilitation exercises (French et al., 2010; Reiman et al., 2012). However, the studies included in both reviews reflect the three major shortcomings of GMed EMG research in general. First, all included studies used surface electrodes. Second, all studies used one electrode to assess the function of the whole muscle. Finally, at least six different electrode placement sites have been described between the studies, therefore each study may potentially be recording myoelectric activity from functionally unique segments of GMed, making it difficult to compare results between studies.

A clearer understanding of the function of GMed is considered essential since GMed is believed to have a major role in lower limb dysfunction (Grimaldi et al., 2009; Müller et al., 2010; Pfirrmann et al., 2005). Hip abductor weakness has been reported in lateral hip pain (Strauss et al., 2010); patello-femoral pain syndrome (PFPS) (Magalhaes et al., 2010;

Nakagawa et al., 2012); osteoarthritis of the hip (Arokoski et al., 2002) and knee (Hinman et al., 2010); and ankle dysfunction (Friel et al., 2006; Kulig et al., 2011).

The aim of this study was therefore to apply recently developed, verified intramuscular EMG guidelines (Semciw et al., 2012a) to determine whether GMed is comprised of functionally independent segments in healthy young adults. This will have implications for our theoretical understanding of the broad function of GMed and may influence future work aimed at assessing the role of GMed in a range of clinical populations.

Methods

1.1 Participants

Fifteen health young adults (9 male, 6 female) volunteered for this study, with a mean (SD) age, height and weight of 22.5 (2.4) years, 177.4 (9.9) cm and 76.9 (12.8) kg respectively. Volunteers were active with an average (SD) of 6.3 (4.4) hours/week of land based exercise and a Tegner Activity Score (Tegner and Lysholm, 1985) of greater than three. Participants were free of hip and lumbar spine disease, pain and injury. This study was approved by the University Human Ethics Committee (UHEC 10-065), and all participants gave informed consent.

1.2 Instrumentation and electrode insertions

Stainless steel, Teflon® coated bi-polar fine wire (A-M Systems, Washington, USA) electrodes were prepared as described by earlier reports (Basmajian and Stecko, 1962; Semciw et al., 2012b). All testing was performed on the stance dominant limb (Bullock-Saxton et al., 2001). Participants were positioned in side lying with their hips and knees in 45° flexion. Anterior, middle and posterior segments of GMed were marked using previously verified guidelines (Semciw et al., 2012a) and real time ultrasound (HDI 3000; Advanced Technology Laboratories, Washington, USA) was used to guide the depth of electrode insertion into the belly of each segment as described previously (Semciw et al., 2012b). A two-inch Dermatrode reference electrode (American Imex, CA, USA) was placed dorsally on the contra-lateral hand. Force sensitive resisters (footswitches) (Model: 402, Interlink Electronics, California, USA) were placed over the heel and great toe to determine the temporal components of the gait cycle (Murley et al., 2009b). Raw signals

from the footswitches, reference electrode and intramuscular electrodes were received by a Delsys® Bagnoli-16 EMG system (Delsys Inc., Boston, USA).

1.3 Experimental protocol

There were two components to the experimental protocol. The first was a series of six walking trials (Murley et al., 2009b) at comfortable self-selected walking speed (Latt et al., 2008) along a 9 m walkway. The last four trials were recorded for analysis, and trials were repeated if they exceeded ± 5% of the average walking speed (established during warm-up).

The second component of the experimental protocol consisted of a series of maximum voluntary isometric contractions (MVICs). It has been recommended that multiple tests be performed in order to obtain the optimum maximum value for a muscle's MVIC (Burden, 2010; Ekstrom et al., 2005; Vera-Garcia et al., 2010), and that a compromise be made on the number of tests performed in order to minimize participant fatigue (Vera-Garcia et al., 2010). Pilot work on eight different positions revealed that external rotation, flexion, and abduction in external rotation were least likely to record a true maximum for any of the GMed segments (Semciw et al., 2011). These three actions were therefore excluded from the testing protocol in this study in order to minimize participant fatigue. MVICs for this study therefore comprised of open chain hip abduction, hip internal rotation, hip abduction in internal rotation, hip extension and the clam exercise. The clam was performed by moving the knees apart against a resistance while keeping feet together in a position of 45º hip and knee flexion (modified from Distefano et al., 2009). All actions were performed in side-lying, except for extension which was performed in prone. The hip remained in the

anatomical position for all actions except the clam. Resistance was applied by a Velcro® strap secured to the plinth and positioned over the participants knee for all actions except internal rotation. Internal rotation was resisted by an investigator, who provided manual resistance at the participants foot while the knee was in 90º of flexion. For each MVIC action, participants were instructed to slowly increase muscle contraction against the resistance, and sustain maximum effort for three seconds. Participants performed three MVIC's for each action and were given a three minute rest in between each contraction. Consistent verbal encouragement was provided by the investigators and the order of MVIC testing was randomly assigned.

1.4 EMG data processing and analysis

Raw EMG signals (Fig. 1A) were passed through a differential amplifier (Delsys Inc., Boston, USA; input impedance = 1015Ω//0.2 pF, CMRR = 92 dB @ 60 Hz) at a gain of 1000, band pass filtered (built into the amplifier) at 20-2000 Hz and sampled at 2000 Hz. To remove low frequency movement artefact, with minimal interruption to the raw EMG signal, a high-pass 4th order Butterworth filter with phase lag was applied (cut-off frequency of 50 Hz) (Chapman et al., 2010). Finally, the signals were full wave rectified and further filtered with a low-pass 4th order Butterworth filter with phase lag, at a cut-off frequency of 6 Hz to generate a linear envelope that would best represent muscle tension through the gait cycle (Murley et al., 2009a; Winter, 1990) (Fig. 1B).

Insert Figure 1 here

Two consecutive strides from the 3rd or 4th stride of each walking trial were further processed for analysis (2 strides x 4 trials = 8 strides per participant) (Murley et al., 2009a). These strides were chosen to ensure participants were not accelerating or decelerating at the point of analysis. For each muscle segment and participant, an ensemble average was generated from the eight strides. All participants ensemble averages were summed and averaged to produce a grand ensemble for GMed anterior, middle and posterior, and establish an EMG profile for each segment across the gait cycle. Consistent bursts of EMG activity were identified in the grand ensemble curve at early stance (0%-20% gait cycle) and mid to late stance (20%-60% gait cycle). Data were therefore acquired from three phases of the gait cycle: 0% to 20%; 20% to 60% and total stance (heel strike to toe-off, 0% to 60% gait cycle). Analysing phases of the gait cycle according to this methodology is consistent with past research where gluteus medius EMG has been analysed in early stance (0% to 20% gait cycle) and mid-stance (20% to 40% gait cycle) (Rutherford and Hubley-Kozey, 2009)

Delsys EMGworks 4.0 signal analysis software was used to acquire the dependant variables from each phase of the gait cycle. These were established from the linear envelopes of each participant's individual trials. For each muscle segment, values were obtained for peak amplitude (%MVIC), average amplitude (%MVIC) and time to peak (TTP, % of gait cycle) from each phase of the gait cycle (0-20%, 20-60%, and total stance).

Data from the five MVIC positions were used for amplitude normalization of gait variables, and for further comparisons between anterior, middle and posterior segmental function. The muscle intensity (RMS amplitude) during an MVIC was calculated from the

middle 1s of each MVIC trial. The highest amplitude value across all five positions was considered MVIC for each segment and for each participant.

The means of amplitude (peak and average) and temporal (TTP) gait variables were compared between muscle segments (anterior, middle and posterior) within each phase of the gait cycle (0% to 20%; 20% to 60%; and total stance) using a one way analysis of variance (ANOVA). Logarithm transformed variables were used where assumptions of normality were not met. Where significant differences were detected ($p<0.05$), post-hoc comparisons were performed with independent samples *t*-test, and a Bonferroni correction was made to account for multiple comparisons (Field, 2009). Significance for post-hoc analysis was therefore set at $\alpha = 0.017$ (0.05 / 3 comparisons). A standardised mean difference (SMD = mean difference / pooled SD) was calculated for all post-hoc comparisons to provide a measure of the magnitude of difference (effect size, ES) between segments (Borenstein et al., 2009), and illustrated with 98% confidence intervals (CI's) to account for Bonferroni adjustments. An ES threshold of 0.2, 0.5 and 0.8 was considered small, medium and large respectively (Cohen, 1988).

The Kruskal-Wallis (K-W) test was used to examine whether GMed segments (x3) were contracting at different relative intensities during each MVIC position (x5). Separate K-W tests were performed for each MVIC using an α of 0.05 to determine significance. Post-hoc comparisons were made with Mann-Whitney U tests ($\alpha = 0.017$, Bonferroni adjustment). A standardised ES was calculated for all post-hoc comparisons by dividing the z-score of the Mann-Whitney U test by the square root of the total sample size (Field, 2009). All statistical comparisons were performed using the SPSS statistical software package (version 19, IBM SPSS Inc., Chicago, IL, USA)

2 Results

All electrode insertions except one remained *in-situ* for the entire testing session. Analysis was therefore conducted on 14 anterior segments, and 15 middle and posterior segments. The mean (±SD) walking speed was 1.17 (0.15) m s^{-1}.

2.1 Gait

The grand ensemble curves demonstrated two consistent bursts of activity for all GMed segments within the stance phase of gait (Fig. 2). There were no significant differences in amplitude variables (peak and average) between segments of GMed (Table 1). However, GMed segments did demonstrate significant differences in TTP for the first ($F_{2,41}$=4.65, p=0.02) and second burst ($F_{2,41}$=6.16, p<0.01) (Table 1). The anterior segments first burst peaked later than the middle segment (p=0.014); and its second burst peaked later than middle and posterior segments (p<0.006) (Fig. 3). These findings were large in magnitude (ES>0.80).

Insert Figure 2 here

Insert Table 1 here

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2.2 MVIC

During MVIC testing (means and SD's available as supplementary data), GMed segments were contracting at significantly different intensities for hip abduction (H_2=8.218,

p=0.016), internal rotation (H_2=24.324, p<0.001), extension (H_2=6.874, p=0.032) and clam (H_2=30.306, p<0.001). No significant difference between segments were apparent during abduction in internal rotation (H_2=3.880, p=0.144). Post-hoc comparisons (Fig. 4) revealed that posterior GMed was contracting at a greater intensity than anterior GMed in abduction (U=168.0, p=0.005); a lower intensity than both other segments in internal rotation (U≥6.0, p<0.001); and a lower intensity than middle GMed during extension (U=51.5, p=0.010). Finally, all pairs of segmental comparisons were significantly different during the clam manoeuvre (U≥187.5, p≤0.001).

Insert Figure 4 here

3 Discussion

This is the first study to characterise EMG profiles of segments of GMed with uniquely oriented muscle fibres using verified electrode insertion guidelines during gait. All three segments display two bursts of activity within the stance phase of gait. The results suggest that three segments are capable of independent function. This was apparent in both the gait and MVIC variables.

3.1 Muscles within muscles

The most convincing evidence of independent functional segments within GMed is demonstrated during a maximum resisted clam manoeuvre (Fig. 4). When segmental muscle activity is classified according to previously described criteria (Reiman et al., 2012), muscle activation during this manoeuvre ranged from a high relative intensity in posterior GMed (mean 47.8% MVIC, refer to supplementary table for segment means) to low intensity in anterior GMed (mean 1.8% MVIC). With moderate to large differences (ES≥0.57) between all segments during a clam exercise, GMed can confidently be described as being composed of muscles within muscles (Wickham and Brown, 1998). This suggests that caution should be used when interpreting prior research that generated conclusions of GMed function based on one electrode recording (French et al., 2010; Reiman et al., 2012).

There is some evidence of segmental activation during the gait cycle as well; however this is restricted to anterior GMed. The peak of anterior GMed occurred later than middle GMed for the first burst and later than both segments in the second burst (Fig 2). This is consistent with the later activity of anterior GMed reported by an earlier fine wire

investigation based on qualitative analysis (Soderberg and Dostal, 1978). Conversely, and in contradiction to prior work (Gottschalk et al., 1989), there was no evidence of phasic activation between posterior and middle GMed during gait. The early activation of posterior relative to middle GMed recorded by Gottschalk et al's surface electrodes is more likely to reflect activity of GMax rather than posterior GMed (Winter and Yack, 1987), particularly given that posterior GMed is completely covered by GMax (Hodges et al., 1997; Semciw et al., 2012a). The current study suggests that middle and posterior segments act synchronously (Fig. 1 and 3), while anterior GMed's first burst peaks later than middle GMed, and the second burst peaks later than both other segments.

3.2 The functional role of gluteus medius during gait

The functional role of each GMed segment across the gait cycle can be viewed in the context of their role as either pelvic stabilisers, or femoral head stabilisers. Muscle segments with a large physiological cross sectional areas (PCSA) and a moment arm in the coronal or transverse plane are better able to generate a high torque for maintaining pelvic equilibrium, or produce pelvic rotation in the transverse plane (on a fixed lower limb) respectively (Neumann, 2010; Ward et al., 2010). Segments with a small PCSA and moment arm, and fibers aligned parallel to the neck of femur (NOF) are likely to contribute to femoral head stability, by drawing the head of femur into the acetabulum (Gottschalk et al., 1989).

Findings of the current study indicate that posterior and middle GMed act synchronously across the gait cycle. However, each segment may serve a different purpose when the current EMG findings are supplemented with biomechanical and morphological muscle

data. The combination of middle GMed's vertical fiber orientation (Al-Hayani, 2009; Gottschalk et al., 1989; Semciw et al., 2012a; Sparks, 2011), large moment arm in the coronal plane (Dostal et al., 1986; Neumann, 2010) and large PCSA (Sparks, 2011) suggest that it has a high potential to generate a large abduction torque in the anatomical position. On a fixed lower limb, this would facilitate pelvic stability. In contrast, posterior GMed has a coronal moment arm, smaller PCSA (Dostal et al., 1986; Neumann, 2010; Sparks, 2011), and its fibers are arranged parallel to the NOF (Al-Hayani, 2009; Gottschalk et al., 1989). These features would facilitate its role as a stabiliser of the head of femur (Al-Hayani, 2009; Gottschalk et al., 1989).

The role of anterior GMed is potentially two-fold. Its relatively vertical fiber orientation (Al-Hayani, 2009; Gottschalk et al., 1989; Semciw et al., 2012a), large moment arm (coronal plane, Dostal et al., 1986; Neumann, 2010) and PCSA (Sparks, 2011) would enable it to assist middle GMed with maintaining pelvic stability across the stance phase of gait (Gottschalk et al., 1989). Additionally, the later peak in EMG activity of anterior GMed during the second burst (late mid-stance) could reflect a supplementary role in two other domains. Anterior GMed may potentially be recruited to minimize anterior hip joint forces associated with an extending hip joint (Lewis et al., 2007), or it may be recruited to contribute to forward contralateral rotation of the pelvis in the transverse plane (Al-Hayani, 2009; Gottschalk et al., 1989; Neumann, 2010). However, given the large PCSA (Sparks, 2011) and favourable moment arm in the transverse plane (Dostal et al., 1986; Neumann, 2010), it is more likely that the latent peak EMG activity (second burst) reflects a contribution to contra-lateral forward rotation of the pelvis (Al-Hayani, 2009; Gottschalk et al., 1989; Neumann, 2010); while anterior GMin (which also has latent peak activity;

Semciw et al., submitted 2013) is better suited morphologically for stabilising the head of femur in mid to late stance.

3.3 Clinical implications

Hip abductor dysfunction has been associated with a range of lower limb disorders (Arokoski et al., 2002; Cowan et al., 2009; Friel et al., 2006; Hinman et al., 2010; Magalhaes et al., 2010; Nakagawa et al., 2012; Strauss et al., 2010). Furthermore, specific localised fatty atrophy has been identified in the anterior third of GMed from 3 to 12 months after a total hip arthroplasty (THA) (Bremer et al., 2011; Müller et al., 2010; Müller et al., 2011). The specific atrophy of anterior GMed in this population is similar to that observed in anterior GMin (Bremer et al., 2011; Pfirrmann et al., 2005), however the potential mechanism may differ. There is kinematic evidence of reductions in peak hip extension range in people following THA (Beaulieu et al., 2010), which may theoretically reduce the stimulus for anterior GMed to provide the rotary torque for contralateral pelvic rotation. Further investigation of segmental function of GMed in clinical populations may determine whether the structural deficits observed in some conditions translate to functional deficits, and what the potential mechanisms of these deficits may be. This knowledge will enable physiotherapists to develop specific and targeted rehabilitation programs for each muscle segment and clinical condition.

3.4 Limitations

The optimal method for normalizing GMed EMG signals recorded from our insertion protocol is unknown. However, a recent review has endorsed the use of MVIC as a normalization method that would enable comparisons between muscles, groups,

interventions or testing conditions in pain free healthy participants (Burden, 2010), and a further study specifically advocates its use in gait normalization (Burden et al., 2003).

It is possible that GMed muscle activation patterns in this study may have been affected by factors such as walking speed and body mass index (Rutherford and Hubley-Kozey, 2009); levels of discomfort associated with the intramuscular electrodes (Henriksen et al., 2009); and lower limb kinematics or kinetics (Beckman and Buchanan, 1995; Bird et al., 2003). However, the mean (± SD) walking velocity of our sample (1.17 ± 0.15 m s^{-1}) is comparable to those reported for other samples of healthy participants ambulating at comfortable walking speed (Latt et al., 2008; Murley et al., 2009a; Rutherford and Hubley-Kozey, 2009), therefore unlikely to be a source of difference between other study populations. Furthermore, the level of discomfort associated with this protocol in this sample of participants was mild (Semciw et al., 2012b), and not considered to significantly affect muscle activity. Finally, altering lower limb biomechanics has influenced GMed muscle activity previously (Beckman and Buchanan, 1995; Bird et al., 2003). It is therefore possible that individual variation in kinematics, kinetics and posture will have influenced EMG recordings of our participants. Further research with concurrent kinematic and kinetic data will be valuable for determining the extent of the association between these variables

4 Conclusion

Validated intramuscular EMG has confirmed that the anterior, middle and posterior segments of GMed have unique functional characteristics. Caution should be used in interpreting results of previous EMG studies of GMed using a single, or only surface electrodes. These results improve the understanding of the function of GMed and pave the way for further research into the role of segments of GMed in clinical populations.

Conflict of interest: none declared

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