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Gait analysis in chronic heart failure: The calf as a locus of impaired walking capacity

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1. Introduction

Chronic heart failure (CHF) is characterized by a marked reduction in walking capacity that in turn contributes to a reduction in quality of life (Juenger et al., 2002). Indeed, preferred walking speed has been reported to be ~30% lower in CHF compared to healthy age-matched control groups (Beneke and Meyer, 1997; Figueiredo et al., 2013) and walking endurance decreases with increasing severity of the disease (Riley et al., 1992). Notably, walking capacity has also been directly linked with hospitalization and mortality rates in CHF (Forman et al., 2012).

Identifying the mechanisms underlying the reduced walking performance in CHF therefore has important functional and clinical relevance. In many instances, skeletal muscle dysfunction, rather than cardiac function, is fundamental to the exercise intolerance evident in this group (Cohn et al., 1993; Cicoira et al., 2001; Fulster et al., 2013). This underlies the “skeletal muscle hypothesis” of exercise intolerance in CHF (Clark et al., 1996). However, most studies addressing the “skeletal muscle hypothesis” in humans have focused primarily on isolated skeletal muscle, including histology, biochemistry (Sullivan et al., 1990), morphology (Anker et al., 1997; Panizzolo et al., 2014) and strength (Upkin et al., 1988; Toth et al., 2006). It is likely that these factors all contribute to some extent to reduced functional walking performance in CHF. Yet, unlike other conditions where skeletal muscle dysfunction purportedly leads to impaired walking, no study to date has examined the detailed biomechanics of walking itself in patients with CHF. Such an analysis would provide
an effective means to identify the end-effect of skeletal muscle dysfunction on walking mechanics in CHF. This information can help reveal the basis for functional limitations in CHF and foster evidence-based rehabilitation approaches aimed at restoring walking capacity.

The goal of this study was, therefore, to perform a detailed biomechanical gait analysis of walking in CHF patients, compared to healthy age-matched control participants. Recently, we identified that the plantarflexors (the triceps surae) undergo proportionately more muscle wasting in CHF than other lower limb muscles (Panizzolo et al., 2014). Moreover, plantarflexor size, unlike the overall leg lean mass, is strongly correlated with peak aerobic capacity of walking in CHF patients (Panizzolo et al., 2014). These characteristics, together with the finding that the plantarflexors are the main source of work during gait in healthy young and old adults (McGowan et al., 2009; De Vita and Hortobagyi, 2000) and that a reduction in walking speed in older adults is related to the triceps surae function (Panizzolo et al., 2013), suggest that restrictions at the ankle joint might particularly affect the ability of CHF patients to achieve the typical gait speed and mechanics seen in a normal healthy population. Accordingly, we hypothesized that (1) a slower walking speed is selected in CHF, compared with healthy age-matched individuals, to reduce total leg mechanical work; (2) the more pronounced wasting reported in plantarflexor muscles in CHF (Panizzolo et al., 2014) requires additional redistribution of mechanical work during stance from the ankle to the other lower limb joints; and (3) this redistribution would be more pronounced at faster walking, where a greater percentage of peak aerobic capacity is utilized. This hypothesis is based on the strong correlation between plantarflexor muscle size and strength and peak aerobic capacity during walking previously reported in CHF (Panizzolo et al., 2014), and the possibility therefore that mechanical work is “shunted” from the ankle to other less aerobically limiting muscles at other joints.

2. Methods

2.1. Subjects

We recruited 10 subjects (6 men, 4 women) with CHF (NYHA class II–IV; ejection fraction 30.9 ± 9.7%, mean ± SD), and 11 healthy subjects from the local community (8 men, 3 women; see Table 1 for subjects characteristics). Exclusion criteria for the CHF population are presented in the Supplementary Material. All subjects were free from musculoskeletal injury and other musculoskeletal diseases and provided written informed consent prior to participating in the study. All procedures were approved by the Human Research Ethics Committee at The University of Western Australia and Royal Perth Hospital.

2.2. Preferred walking speed

A protocol based on over-ground and treadmill walking trials was used to define each subjects’ preferred treadmill walking speed. The detailed description of this protocol is presented in Panizzolo et al. (2013) and in the Supplementary Material.

2.3. Joint and lower limb mechanical work and cost of transport (COT)

Biomechanical measurements were collected with subjects wearing their own sport shoes while walking on an instrumented split-belt treadmill measuring 3D ground reaction forces (Bertec, Columbus, OH, USA; 2000 Hz) at three different walking speeds: the subjects’ preferred speed, a speed 20% faster than their preferred speed and a speed 20% slower than their preferred speed. Subjects walked at each testing speed for approximately 30 to 1 minute before data collection.

Three-dimensional (3D) gait analysis was performed on each subject during their treadmill walking trials. The marker set and configuration used for 3D motion capture (VICOM, Oxford Metrics, UK; 100 Hz) was similar to that of Besier et al. (2003), with the addition of torso markers (a full description of the marker set is provided in the Supplementary Material). Functional joint centers for the hip and knee were defined using the procedures of Besier et al. (2003); All markers and force trajectories were filtered using a zero-lag fourth-order low-pass Butterworth filter with a 5–7 Hz optimal cut-off frequency that was selected using a custom residual analysis algorithm (MATLAB, The MathWorks Inc., USA).

Marker positions collected during a static trial were used to generate subject-specific musculoskeletal models in OpenSim 2.0.2 (Delp et al., 2007). The generic OpenSim musculoskeletal model (Arnold et al., 2010) was scaled using an inverse kinematics algorithm based on the position of the markers placed on anatomical landmarks and on the functional joint centers previously determined. Joint angles, net moments and instantaneous powers were computed using inverse kinematics and inverse dynamics performed in the joint coordinate systems of the scaled model for the walking trials. These calculations were made directly in OpenSim by combining 3D markers trajectories and measured ground reaction forces. Positive joint work was computed for the stance phase and for the complete stride by integrating the positive values of the instantaneous joint power traces (further details of the joint mechanical measurements are provided in the Supplementary Material). The total positive work in the lower limbs was computed both for the stance phase and for the entire stride from the sum of each joint (left and right legs were computed individually and summed) and normalized to the lower limb lean mass (see below). The distribution of total work between individual joints was computed by dividing the total lower limb work by the positive work in the individual joints (sum of left and right joints). Total joint work was normalized to the triceps surae volume. We analyzed both the mechanical work performed during the stride and stance phase, as well as the mechanical cost of transport (work per distance travelled). Main effects of group and speed and interaction effects were evaluated in these analyses. The distribution of work produced across joints was evaluated with a 3 × 2 mixed model repeated measures ANOVA including joint kinematics (minimum, maximum and total angle range) and kinetics (peak moments), the positive work in the combined lower limb (normalized by lower limb lean mass) and the plantarflexor work at the ankle normalized to the triceps surae volume. We analyzed both the mechanical work performed during the stride and stance phase, as well as the mechanical cost of transport (work per distance travelled). Main effects of group and speed and interaction effects were evaluated in these analyses. The distribution of work produced across joints was evaluated with a 3 × 2 MANOVA (significant level of p < 0.05) including the three lower limb joints. An analysis of covariance (ANCOVA) was applied comparing the results of this analysis. Joint and speed were set as multivariate factors and group as a univariate factor. Where significant main and/or interaction effects were detected in ANOVA or MANOVA analyses a Bonferroni correction was applied to the percentage joint contribution prior to performing these analyses. The distribution of work produced across joints was evaluated with a 3 × 2 mixed model repeated measures ANOVA including joint kinematics (minimum, maximum and total angle range) and kinetics (peak moments), the positive work in the combined lower limb (normalized by lower limb lean mass) and the plantarflexor work at the ankle normalized to the triceps surae volume. We analyzed both the mechanical work performed during the stride and stance phase, as well as the mechanical cost of transport (work per distance travelled). Main effects of group and speed and interaction effects were evaluated in these analyses. The distribution of work produced across joints was evaluated with a 3 × 2 MANOVA (significant level of p < 0.05) including the three lower limb joints. An analysis of covariance (ANCOVA) was applied comparing the results of this analysis. Joint and speed were set as multivariate factors and group as a univariate factor. Where significant main and/or interaction effects were detected in ANOVA or MANOVA analyses a Bonferroni post hoc test was conducted.

Table 1

Subject characteristics. Data are means ± SD.

<table>
<thead>
<tr>
<th>Group</th>
<th>Age [yr]</th>
<th>Height [m]</th>
<th>Weight [kg]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Control</td>
<td>63.1 ± 5.6</td>
<td>1.73 ± 0.06</td>
<td>70.1 ± 8.8</td>
</tr>
<tr>
<td>CHF</td>
<td>60.7 ± 9.8</td>
<td>1.67 ± 0.10</td>
<td>73.0 ± 19.0</td>
</tr>
</tbody>
</table>

The chronic heart failure (CHF) group underwent regular exercise activity 6 times per week for an average of 6 months. The detailed description of this protocol is presented in Panizzolo et al. (2013) and in the Supplementary Material. The total positive work in the lower limbs was computed both for the stance phase and for the entire stride from the sum of each joint (left and right legs were computed individually and summed) and normalized to the lower limb lean mass (see below). The distribution of total work between individual joints was computed by dividing the total lower limb work by the positive work in the individual joints (sum of left and right joints). Joint and speed were set as multivariate factors and group as a univariate factor. Where significant main and/or interaction effects were detected in ANOVA or MANOVA analyses a Bonferroni correction was applied to the percentage joint contribution prior to performing these analyses. The distribution of work produced across joints was evaluated with a 3 × 2 mixed model repeated measures ANOVA including joint kinematics (minimum, maximum and total angle range) and kinetics (peak moments), the positive work in the combined lower limb (normalized by lower limb lean mass) and the plantarflexor work at the ankle normalized to the triceps surae volume. We analyzed both the mechanical work performed during the stride and stance phase, as well as the mechanical cost of transport (work per distance travelled). Main effects of group and speed and interaction effects were evaluated in these analyses. The distribution of work produced across joints was evaluated with a 3 × 2 MANOVA (significant level of p < 0.05) including the three lower limb joints. An analysis of covariance (ANCOVA) was applied comparing the results of this analysis. Joint and speed were set as multivariate factors and group as a univariate factor. Where significant main and/or interaction effects were detected in ANOVA or MANOVA analyses a Bonferroni post hoc test was conducted.

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3. Results

3.1. Preferred walking speed and spatio-temporal parameters

Preferred walking speed on the treadmill and over-ground were not significantly different between groups. Spatial-temporal gait parameters including stride length, stride frequency, duty factor, stance and swing times were similarly not significantly different between groups (Table 2).

3.2. Joint kinematics and kinetics

No differences were found between CHF and control participants in peak joint moments or powers at the ankle, knee or hip across all three speeds. A main effect of group was reported in the kinematics of the ankle. The CHF group exhibited a higher dorsiflexion peak in stance (p=0.01) but no differences in the total ankle range of motion (p=0.2). Joint kinematics, kinetics and power traces for the preferred speed are displayed in Fig. 1.

3.3. Total lower limb lean mass-specific and plantarflexor volume-specific work and COT

Total leg lean mass-specific mechanical work and COT across the stride were on average 10.2% and 15.6% greater in CHF versus control (Fig. 2), although no statistical main effect of group was observed in either of these parameters (p=0.2 and p=0.1, respectively). A significant main effect of speed was present in both lower limb lean

Table 2
Spatio-temporal parameters at different testing speeds. Data are means ± SD.

<table>
<thead>
<tr>
<th>Speed</th>
<th>Group</th>
<th>Testing speed [m/s]</th>
<th>Stance time [s]</th>
<th>Swing time [s]</th>
<th>Duty factor</th>
<th>Stride freq. [Hz]</th>
<th>Stride length [m]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Slow</td>
<td>Control</td>
<td>0.81 ± 0.13</td>
<td>0.77 ± 0.07</td>
<td>0.51 ± 0.06</td>
<td>0.60 ± 0.01</td>
<td>0.79 ± 0.08</td>
<td>1.03 ± 0.14</td>
</tr>
<tr>
<td></td>
<td>CHF</td>
<td>0.79 ± 0.20</td>
<td>0.80 ± 0.12</td>
<td>0.51 ± 0.07</td>
<td>0.61 ± 0.03</td>
<td>0.78 ± 0.10</td>
<td>0.96 ± 0.21</td>
</tr>
<tr>
<td>Preferred</td>
<td>Control</td>
<td>1.00 ± 0.16</td>
<td>0.69 ± 0.06</td>
<td>0.48 ± 0.05</td>
<td>0.59 ± 0.01</td>
<td>0.87 ± 0.08</td>
<td>1.16 ± 0.16</td>
</tr>
<tr>
<td></td>
<td>CHF</td>
<td>0.99 ± 0.26</td>
<td>0.74 ± 0.13</td>
<td>0.48 ± 0.06</td>
<td>0.61 ± 0.02</td>
<td>0.84 ± 0.13</td>
<td>1.11 ± 0.22</td>
</tr>
<tr>
<td>Fast</td>
<td>Control</td>
<td>1.21 ± 0.21</td>
<td>0.64 ± 0.05</td>
<td>0.45 ± 0.04</td>
<td>0.59 ± 0.01</td>
<td>0.93 ± 0.07</td>
<td>1.28 ± 0.18</td>
</tr>
<tr>
<td></td>
<td>CHF</td>
<td>1.19 ± 0.31</td>
<td>0.67 ± 0.12</td>
<td>0.46 ± 0.07</td>
<td>0.59 ± 0.02</td>
<td>0.91 ± 0.14</td>
<td>1.23 ± 0.24</td>
</tr>
</tbody>
</table>

Fig. 1. Comparison of joint angles, moments and powers (top to bottom) for the preferred walking speed across the gait cycle in chronic heart failure (solid lines) and control subjects (dotted lines): (a) ankle, (b) knee and (c) hip. Data are group means; the shaded regions represent the SD of the mean. The vertical lines represent toe off (solid lines represent the CHF group and the dotted line the control group); Positive joint angles represent flexion (dorsi-flexion at the ankle) and negative angles represented extension (plantarflexion at the ankle). Positive moments represent net flexion joint moments (dorsi-flexion at ankle) and negative moments represent extension joint moments (plantarflexion at ankle). Positive powers represent instantaneous joint power generation and negative powers represent instantaneous joint power absorption. The * represents statistically different ankle joint dorsiflexion in stance (p < 0.05).
mass-specific total work \((p < 0.001)\) and COT \((p < 0.001)\). Post hoc tests revealed that when collapsed across groups differences were present between all speeds for lower limb lean mass-specific total work. Post hoc tests also revealed that the slowest and preferred speeds did not differ in lower limb lean mass-specific COT, but that each was lower than the COT at the fastest speed. No significant interaction effects for positive leg lean mass-specific work or COT were present. The same statistical findings existed for the total leg work when normalized to body mass.

A main effect of group on the triceps surae volume-specific plantarflexion work and COT in stance was observed (Fig. 3); \(p = 0.024\) and \(p = 0.008\), respectively. The amount of work produced by the CHF group was 68.4%, 74.0% and 54.4% higher than the control group (for the three testing speeds); the COT was 68.4%, 74.0% and 54.4% higher than the control group (for the three testing speeds). A main effect of speed was observed on the triceps surae volume-specific plantarflexion work and COT in stance \((p < 0.001\) and \(p = 0.002\), respectively). Post hoc tests revealed that when collapsed across groups the triceps surae volume-specific plantarflexion work were different from each other between all speeds and that the COT at the slowest and preferred speeds were not different from each other but both were lower than that at the fastest speed. No interaction effect was observed between group and speed \((p = 0.2\) and \(p = 0.3\), respectively).

3.4. Distribution of joint work

Across all the testing speeds the hip produced the highest amount of positive work, while the knee joint produced the least work, in both control and CHF groups. Nevertheless, a different distribution of work during stance was found between the CHF and the control group (Fig. 4). Main multivariate effects of group and speed on work distribution were not present \((p = 0.3\) and \(p = 0.08\), respectively) but an interaction effect was found between joint and group \((p = 0.01)\) and between speed and joint \((p = 0.005)\). Post hoc analyses revealed differences in the percent work between groups at the ankle at all speeds \((p = 0.011, p = 0.023,\) and \(p = 0.001)\) and at the hip only at the slowest speed \((p = 0.044, p = 0.13\) and \(p = 0.055)\). These results reflected that the percentage of work produced at the ankle joint was higher in the CHF, while an opposite trend was found for the hip, and that the ankle produced proportionately more work in CHF vs. control participants as speed increased (Fig. 4).

3.5. Aerobic capacity

\(\dot{V}O_2\) peak was significantly higher \((p<0.001)\) in the control group than in CHF \((35.8 \pm 7.9\) and \(15.8 \pm 2.8 \text{ ml/kg/min}, \text{ respectively})\). While no differences were reported in the sub-maximal oxygen consumption values between groups at any of the testing speeds.
4. Discussion

4.1. Speed and overall gait mechanics

One reason why CHF patients may have previously exhibited reduced walking function, including speed, is to compensate for lower limb muscle mechanical capacities (Clark et al., 1996; Harrington et al., 1997; Toth et al., 2006). Our first hypothesis that a reduction in preferred speed is present and is associated with equal levels of mechanical work compared to a faster walking healthy control group was, however, not met. Surprisingly, the preferred speed, both over-ground and on the treadmill, were not statistically different between groups, nor were the spatio-temporal gait parameters. Interestingly, the similarity in gait function between the CHF and control groups extended to joint kinetics and kinematics, with few clearly discernable differences in either joint angles, net joint torques or power with the exception of a more dorsiﬂexed ankle joint in CHF patients (Fig. 1). One possible explanation for the lack of difference in preferred speed in our study compared to others (Beneke and Meyer, 1997; Figueiredo et al., 2013) might be the training status of our patients. Because our CHF patients underwent moderate levels of treadmill exercise it is possible that their similar walking speed was due to a training effect (Beneke and Meyer, 1997). Thus, despite the CHF patients presenting with severely limited \( VO_2 \text{peak} \), regular training may possibly promote similar submaximal walking patterns. It remains possible, therefore, that more pronounced differences in preferred speed and joint mechanics would be detected if CHF patients without training were analyzed.

Another, associated reason for our lack of difference in walking speed, versus previous studies, is the disease severity. If previous studies included subjects with more severe CHF (e.g., NYHA III-IV), they may have been less capable of achieving “normal,” albeit compensated, walking speeds.

4.2. Relative muscular work and distribution of joint work

The similarity in speed and joint mechanics at ﬁrst hand suggest that there are no major biomechanical gait differences present in CHF patients. However, a second possible strategy to offset the

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influence both the acute effort during the stance phase (J/ml over the stance, Fig. 3a) as well as the effort to travel a given distance (J/ml/m Fig. 3b). These factors may be a primary cause of the earlier symptoms of walking fatigue characteristic of this population (Clark et al., 1996), and may be an important limitation to walking aerobic capacity. It is intriguing to consider whether the higher dorsiflexion angles during stance in CHF (Fig. 1) also lead to altered in vivo muscle mechanics. Rubenson et al. (2012) reported operating lengths of the soleus in healthy young adults that are conserved to the ascending limb of the force-length curve during gait. The higher dorsiflexion angles in CHF may lead to longer muscle lengths that generate passive forces, thus partly compensating for the reduced active force capacity of the muscle (although possibly at the increased risk of muscle strain injury). At present this mechanisms remains speculative and requires further in vivo analyses to test.

Surprisingly, the greater demand on the ankle to produce the limb mechanical work increased, rather than decreased, with speed (Figs. 3a and 4), further taxing the small triceps surae muscle mass. Our earlier work on the triceps surae in CHF established a strong link between muscle size and VO₂ peak (Panizzolo et al., 2014), leading to our prediction that mechanical work would be “shunted” away from the ankle as CHF patients walk closer to their aerobic capacity, thus potentially shifting reliance on to other less aerobically limiting muscle groups. The finding that the opposite occurs may in fact help to explain the strong link between the size of the triceps surae and VO₂ peak during walking. Indeed, these muscles were key in powering walking, compared to the control group that relied more on the hip, and were responsible for the majority of mechanical work during fast walking.

4.3. What dictates preferred walking speed and joint work modulation in CHF?

Our results offer some intriguing questions about what factors influence the self-selected walking mechanics in CHF. Firstly, why did the CHF patients not walk slower despite their reduction in muscle size and impaired aerobic capacity? A possible explanation may be an optimization of speed relative to the mechanical COT. We found that the mechanical work required to travel a given distance was not reduced when shifting from the preferred speed to a slower walking speed in either group (Fig. 2b). On the other hand, the COT was elevated in both groups when walking faster than the preferred speed. In this regard, the self-selected speed may represent the fastest speed possible before increasing the mechanical COT. This is plausible since the cumulative (repetitive) loading affects fatigue and energy use in the musculature. Mechanical gait optimization may represent an alternative, or possibly parallel, determinant of gait speed to the respiratory optimization recently reported in CHF patients (Figueiredo et al., 2013).

Secondly, why do CHF patients rely on the ankle to a greater extent to power walking compared to healthy age-matched adults? It has been established previously that aging causes an increased reliance on the hip to power walking (DeVita and Hortobagyi, 2000; Silder et al., 2008). Our data on healthy older adults supports this, with over 50% of the work attributed to the hip in this group (Fig. 4). It is not entirely clear why, in the CHF group, the contributions of each joint to the total mechanical work resemble those in healthy young adults (Farris and Sawicki, 2012), especially considering the reduced muscle volume of the ankle plantarflexors. A first possible explanation might be found in the hip flexor function. These muscle groups have been found to be weaker in CHF (Toth et al., 2006; Toth et al., 2010; Minotti et al., 1993), or functionally impaired. On the other hand, previous studies (Hartridge et al., 1996; Carrington et al., 2001; Panizzolo et al., 2014) indicate that the strength of ankle plantarflexors, despite their prominent loss of muscle volume, may...
be less affected in CHF compared to hip muscle strength. Thus, a greater reliance on the ankle may reflect a more pronounced loss of the muscle force capacity at the hip in CHF compared to healthy older adults.

4.4. Summary

The preferred speed and overall joint kinematics and kinetics are similar between CHF and age- and exercise level-matched subjects. Nevertheless, a marked increase in the muscle volume-specific plantarflexion work and a greater reliance on the ankle over other joints to power walking in CHF patients may help explain their decreased walking capacity. The present study strengthens the finding that the plantarflexor muscles (triceps surae) are a key muscle group limiting exercise capacity (Panizzolo et al., 2014) and should be included in the design of exercise-based rehabilitation specific to this population.

Conflict of Interest

The authors declare that no conflict of interest exists with any of the authors of this article.

Uncited reference

Treece et al., (2003).

Acknowledgments

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Appendix A. Supporting information

Supplementary data associated with this article can be found in the online version at http://dx.doi.org/10.1016/j.jbiomech.2014.09.015.

References


