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

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ABSTRACT

Reduced walking capacity, a hallmark of chronic heart failure (CHF), is strongly correlated with hospitalization and morbidity. The aim of this work was to perform a detailed biomechanical gait analysis to better identify mechanisms underlying reduced walking capacity in CHF. Inverse dynamic analyses were conducted in CHF patients and age- and exercise level-matched control subjects on an instrumented treadmill at self-selected treadmill walking speeds and at speeds representing +20% and –20% of the subjects' preferred speed. Surprisingly, no difference in preferred speed was observed between groups, possibly explained by an optimization of the mechanical cost of transport in both groups (the mechanical cost to travel a given distance; J/kg/m). The majority of limb kinematics and kinetics were also similar between groups, with the exception of greater ankle dorsiflexion angles during stance in CHF. Nevertheless, over two times greater ankle plantarflexion work during stance and per distance traveled is required for a given triceps surae muscle volume in CHF patients. This, together with a greater reliance on the ankle compared to the hip to power walking in CHF patients, especially at faster speeds, may contribute to the earlier onset of fatigue in CHF patients. This observation also helps explain the high correlation between triceps surae muscle volume and exercise capacity that has previously been reported in CHF. Considering the key role played by the plantarflexors in powering walking and their association with exercise capacity, our findings strongly suggest that exercise-based rehabilitation in CHF should not omit the ankle muscle group.

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; Fülster 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 (Lipkin 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

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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 plantarflexor muscles (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 later hypothesis is based on the strong correlation between both 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 ± S.D.) 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 s to 1 min 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 (VICON, 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 positive lower limb work by the positive work in the individual joints (sum of left and right joints). The total mechanical cost of transport (COT) J/kg/m was calculated by dividing the lower limb lean mass-normalized positive work over the stride by the distance traveled over the stride.

The work produced in plantarflexion during stance was computed separately and normalized to the triceps surae volume (see below). Similarly, the specific COT for ankle plantarflexion was computed as the sum of the left and right triceps surae volume-specific positive ankle plantarflexion joint work divided by the distance traveled over one stride. Five non-consecutive strides per speed were used for generating mean kinematic, kinetic and work data for each individual subject, which were subsequently combined to calculate group mean data.

2.4. Lean lower limb mass (DXA) and plantarflexor volume (3Dultrasound)

Overall body composition was determined using DXA (Luna Prodigy, EnCore 2004, GE Medical Systems, Madison, WI, USA) on each subject. Lean mass for the lower limbs was computed separately by selecting a region of interest from the great trochanter to the pubic symphysis and including the leg and foot.

Plantarflexor volume was computed using a three-dimensional ultrasound technique (3DUS) based on a combination of B-mode ultrasound imaging and 3D motion data, the procedures of which have been described in detail in Barber et al. (2009) and Panizzolo et al. (2014) and are reported in the Supplementary Material.

2.5. Peak and sub-maximal oxygen consumption

On a separate testing day, peak oxygen uptake ($\dot{V}O_2$ *peak*) was assessed using an incremental walking protocol on a motorized treadmill (Panizzolo et al., 2013). Indirect calorimetry was conducted using a Vmax Encore gas analysis system (Sensormedics, Yorba Linda, California), which enabled the measurement of expired gas concentrations and volumes. Absolute $\dot{V}O_2$ *peak* was expressed in ml/min and normalized to body mass (ml/kg/min). Sub-maximal oxygen consumption at the three subject-specific speeds described above were also obtained.

2.6. Statistical analysis

A 3×2 mixed model repeated measures ANOVA was performed to evaluate differences between the control and CHF groups across the three testing speeds. The significance level was set at $p < 0.05$ (SPSS Inc., Statistics 21, USA). ANOVA analyses included joint kinematics (minimum, maximum and total angle range) and kinetics (peak moments), the positive work in the combined lower limbs (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 arcsin conversion was applied to the percentage joint contribution prior to performing 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.

Group	Age [yr]	Height [m]	Weight [kg]
Control	63.1 ± 5.6	1.73 ± 0.06	70.1 ± 8.8
CHF	60.7 ± 9.8	1.67 ± 0.10	73.0 ± 19.0

The chronic heart failure (CHF) group underwent regular exercise activity 2–3 times per week for ~1 hour per session (treadmill walking and resistance weight training) for a duration of 3.8 ± 1.3 months as part of their standard patient care. The control subjects underwent similar levels of weekly exercise.