Muscle Activation and Estimated Relative Joint Force During Running with Weight Support on a Lower-Body Positive-Pressure Treadmill

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Running on a lower-body positive-pressure (LBPP) treadmill allows effects of weight support on leg muscle activation to be assessed systematically, and has the potential to facilitate rehabilitation and prevent overloading. The aim was to study the effect of running with weight support on leg muscle activation and to estimate relative knee and ankle joint forces. Runners performed 6-min running sessions at 2.22 m/s and 3.33 m/s, at 100%, 80%, 60%, 40%, and 20% body weight (BW). Surface electromyography, ground reaction force, and running characteristics were measured. Relative knee and ankle joint forces were estimated. Leg muscles responded differently to unweighting during running, reflecting different relative contribution to propulsion and antigravity forces. At 20% BW, knee extensor EMG_{peak} decreased to 22% at 2.22 m/s and 28% at 3.33 m/s of 100% BW values. Plantar flexors decreased to 52% and 58% at 20% BW, while activity of biceps femoris muscle remained unchanged. Unweighting with LBPP reduced estimated joint force significantly although less than proportional to the degree of weight support (ankle). It was concluded that leg muscle activation adapted to the new biomechanical environment, and the effect of unweighting on estimated knee force was more pronounced than on ankle force.

Keywords: rehabilitation, body weight support, EMG, estimated joint load, running

Early mobilization after injury and surgery on the lower extremities is essential to reduce adverse effects of immobilization and to facilitate rehabilitation. Walking or running with weight support on a lower-body positive-pressure (LBPP) treadmill is a new training and rehabilitation modality.^{1,2} A recent study indicated strong correlation between tibiofemoral force and treadmill ground reaction force in a small group of elderly patients during slow walking.³ Another study showed that runners can reduce peak vertical ground reaction forces while maintaining the aerobic stimulus when running on a LBPP treadmill.⁴ Even training at high speeds and high aerobic stimuli with the benefit of low vertical ground reaction force and near-normal movement pattern can be performed on the LBPP treadmill.⁵ Furthermore, no adverse impact on systemic and head cardiovascular parameters has been found.⁶ However, the effect of different levels of LBPP weight support during running on neuromuscular adaptations and their possible associations to joint forces have not been studied systematically.

During level running, leg muscles support body weight (BW) against gravity and accelerate/decelerate the body mass in a forward direction. BW, as measured in Newton, is directly related to gravity, while body mass (kg) is independent of gravity. Therefore, by unweighting in a LBPP chamber, it is possible to create a new

biomechanical environment where the vertical forces are reduced while forces associated with horizontal acceleration/deceleration of the body remain unchanged. This allows systematic study of individual leg muscle contribution to body support against gravity and to propulsion. It seems reasonable to expect a muscle-specific effect of gradual weight support on muscle activation during running in a LBPP chamber. However, knowledge is sparse regarding the muscle-specific effect of increasing weight support on muscle activation levels and temporal activation patterns during running.

Lower extremity joint compression forces during running have been assessed using electromyography (EMG) based models, inverse dynamics, and model simulations assuming a priori force distributions. Values between 8 to 14 times BW for knee and ankle joint compression forces during running have been reported in the literature.⁷⁻¹⁰ The relative large variations between studies are attributed to the different models and assumptions used in the studies. Ground reaction forces as well as muscle forces contribute to joint forces, with muscle forces being the major contributor to mechanical joint load.⁹ It has been estimated that muscle forces amount to 7 to 8 times BW, while ground reaction force contributes an additional 2 to 3 times BW during running.¹⁰ We hypothesized that vertical ground reaction force is reduced proportionally to increasing weight support, while muscle activation is reduced less than proportional to the degree of weight support, as forces related to horizontal propulsion remain unchanged despite changes in weight support. Furthermore, we hypothesized that estimated joint force is reduced markedly but less than proportional to the degree of weight support. The main aim was to study the effect of weight support with LBPP on leg muscle activation, ground reaction force, and estimated relative knee and ankle joint forces during running. In addition, effects of weight support on running characteristics were assessed.

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Methods

Participants

Eight healthy, experienced male runners with a mean age of 27.6 (± 2.4) years, body mass 80.1 (± 6.8) kg, and body height 1.84 (± 0.07) m participated. Resting blood pressure and heart rate (sitting position) were 140/77 ($\pm 9.4/\pm 7.0$) mmHg and 63 (± 8) bpm, respectively. All subjects gave written consent to participate in the study, which was approved by the local ethics committee.

Protocol

The subjects performed 2 running sessions at 2.22 m/s and 3.33 m/s on a force treadmill with a LBPP chamber (Alter-G P200, Alter G, California, USA), which allowed BW to be reduced gradually from 100% BW to 20% BW. The LBPP treadmill system was constructed as a treadmill (Woodway, USA) surrounded by a pressure chamber. The subjects wore a special type of shorts and were zipped at the waist into a pressurized airtight enclosure that is suspended over the surface of the treadmill. By controlling the pressure in the enclosure, it is possible to support BW of an individual to 20% of normal BW in 1% increments. The treadmill is mounted with 4 force transducers (aluminum load cell, 1000 LBF, 2.0 mV/V), which allowed vertical ground reaction force during ambulation to be monitored. The pressure/ground reaction force relation in the LBPP chamber was calibrated during still standing for each participant individually. The system allows unrestricted mobility and natural movement pattern during ambulation. The 2 running sessions were performed in random order and at least 20 minutes of rest was allowed between sessions to avoid fatigue. Each session consisted of 6-minute running trials at 100% BW, 80% BW, 60% BW, 40% BW, and 20% BW. At all levels of BW support, subjects were instructed to run as normal as possible with the body centered in the pressure chamber.

Measurements and Data Analysis

Bipolar surface EMG (8-channel, Logger Technology, Sweden) was recorded from the vastus lateralis (VL), vastus medialis (VM), biceps femoris (BF), lateral gastrocnemius (GAS), and lateral soleus (SOL) muscles using Ag/AgCl electrodes (Neuroline 720, Ambu, Denmark) that were placed with an interelectrode center-to-center distance of 2 cm. The skin was carefully cleaned with alcohol, shaved, and rubbed to obtain an interelectrode resistance below 10 K Ω . EMG was recorded during running and during maximum contractions performed before the running sessions. Four types of maximum contractions were performed. Each type was repeated 3 times. Maximum EMG (EMG_{max}) for the VL and VM muscles was obtained during isometric knee extensions performed in sitting position with a knee angle of 100° degrees. Maximum isometric knee flexions were performed in the same position to obtain maximum EMG from the BF muscle. Maximum EMG for the SOL and GAS muscles was recorded during maximum isometric ankle plantar flexions measured with the subjects in a sitting position (90° knee angle, 90° ankle angle) with a vertical lower leg. The leg was fixed to the floor with a strap across the knee. Finally, maximum plantar flexions were performed while the subjects were standing on 1 leg, while forces were exerted on the shoulders. EMG was recorded using a sampling frequency of 1000 Hz (DT BNC Box USB 9800 Series, 16 bit, Data translation Inc., USA). Root mean square (RMS) amplitudes of the EMG were calculated in 100-millisecond

segments. For the maximum contractions, moving averages of 10 consecutive RMS values (1 s) were calculated. The largest 1-second value for each muscle was determined as maximum EMG (EMG_{max}) and used to normalize EMG measured during running (%EMG_{max}) to allow comparisons of relative EMG between muscles. Average muscle activity (EMG_{mean}) and peak muscle activity (EMG_{peak}) (the 90% percentile, ie, the level exceeded 10% of the time) during running were calculated for each trial in minutes 4 to 6. EMG_{peak} is expected to reflect peak force in the stance phase.

Treadmill force transducers were calibrated using 8 loads between 0 kg and 71 kg. The static calibration showed a linear relation between external load and transducer output ($R^2 = .99$). The dynamic damping characteristics based on single impacts between 900 N and 2400 N showed a resonant (fundamental) frequency of 16.2 Hz and an impact amplitude reduction of 94% at T0. Thus, the transducers were high precision load cells with low resonance. The summed force signal from the 4 load cells in the treadmill was recorded during the running sessions with a frequency of 1000 Hz (DT BNC Box USB 9800 Series, 16 bit, Data translation Inc., USA). Force data were used to calculate peak ground reaction forces (GRF_{peak}) as the 90% force (ie, the level of force exceeded 10% of the time) and mean ground reaction forces (GRF_{mean}). Force calculations are based on minutes 4 to 6 during the trials. Furthermore, measures of running kinematics (cadence and relative stance duration) were calculated based on GRF data. The reported data are average values for the fifth minute of the trials.

Model Estimation of Relative Joint Force

Relative knee and ankle peak joint force reductions at weight-supported conditions were estimated based on GRF_{peak} and EMG_{peak} values and assuming 20% contribution from ground reaction force and 80% contribution of muscle activation to joint forces:¹⁰ relative peak joint force = (k_1 (rel - GRF_{peak}) + k_2 (rel - EMG_{peak})) $\times$ 100% BW, where $k_1 = 0.2$, $k_2 = 0.8$, and rel - GRF_{peak} = GRF_{peak} at weight-supported condition divided by GRF_{peak} at 100% BW. For the knee: rel - EMG_{peak-k} = (EMG_{peak} for VL + EMG_{peak} for VM) / 2 at weight-supported conditions divided by (EMG_{peak} for VL + EMG_{peak} for VM) / 2 at 100% BW.

For the ankle: rel - EMG_{peak-a} = (EMG_{peak} for GAS + EMG_{peak} for SOL) / 2 at weight supported conditions divided by (EMG_{peak} for GAS + EMG_{peak} for SOL) / 2 at 100% BW.

Statistical Analysis

Statistical analyses were performed using Minitab (MTB version 13; Minitab, USA). A general linear model with BW and velocity as fixed factors was used to identify the effect of weight support and velocity on EMG_{mean}, EMG_{peak}, GRF_{mean}, GRF_{peak}, step frequency, and relative duration of the stance phase. No interactions were found between BW and velocity. Estimated knee joint load and estimated ankle joint load were compared by paired *T*-test. The threshold for statistical significance was set at $P = .05$.

Results

The time courses of the muscle activation patterns at different levels of weight support were qualitatively similar, although leg muscles responded differently regarding amplitude adaptations to gradual weight support. Muscle activation pattern of the leg muscles as well as ground reaction force for a typical subject are shown in Figure 1.

Figure 1 — An example of vertical ground reaction force (GRF) and raw electromyography (EMG) from the vastus lateralis (m.vl), vastus medialis (m.vm), biceps femoris (m.bf), gastrocnemius (m.gas), and the soleus (m.sol) muscles while running at 100% body weight (BW), 60% BW, and 20% BW ($n = 1$). A major response to reduced BW was seen for the knee extensor muscles. A less marked response was seen for the plantar flexor muscles.

EMG_{mean} (VL, VM, SOL, GAS) decreased with increasing weight support ($P < .001$) (Figure 2A, 2B). Increasing weight support from 100% BW to 20% BW decreased EMG_{mean} to 24.8% (VL) and 23.3% (VM) of the values measured at 100% BW at 2.22 m/s, and to 36.1% (VL) and 28.4% (VM) at 3.33 m/s. The corresponding values for the GAS and SOL muscles were 49.9% and 46.5% of 100% BW values at 2.22 m/s, and 58.8% and 57.7% at 3.33 m/s. EMG_{mean} of the BF muscle remained unchanged ($P = .959$) (overall average 9.3% EMG_{max} at 2.22 m/s and 10.8% EMG_{max} at 3.33 m/s) when BW was increasingly supported.

EMG_{peak} (VL, VM, SOL, GAS) decreased with increasing BW support ($P < .001$) (Figure 2C, 2D). Thus, unweighting from 100% BW to 20% BW at 2.22 m/s decreased EMG_{peak} to 23.2% (VL) and 19.8% (VM) of the initial values measured at 100% BW and to 29.7% (VL) and 26.2% (VM) at 3.33 m/s. For the GAS and the SOL muscles, the corresponding values were 53.8% and 50.6% of 100% BW values at 2.22 m/s, and 58.1% and 58.6% at 3.33 m/s. EMG_{peak} of the BF muscle remained unchanged ($P = .754$) (overall mean 19.3% EMG_{max} at 2.22 m/s and 23.5% EMG_{max} at 3.33 m/s) despite reduced BW. Knee extensor EMG_{peak} (mean of VL and VM muscles) increased by 5.0% from 2.22 m/s to 3.33 m/s, and ankle plantar flexor EMG_{peak} (mean of SOL and GAS) increased by 9.5% from 2.22 m/s to 3.33 m/s across all BWs.

GRF_{mean} decreased from 98% BW to 20% BW at 2.22 m/s, and from 99% BW to 23% BW at 3.33 m/s when unweighting from 100% BW to 20% BW ($P < .001$). GRF_{peak} decreased from 176% BW to 55% BW at 2.22 m/s and from 191% BW to 68% BW at 3.33 m/s when unweighting from 100% BW to 20% BW ($P < .001$) (Figure 3).

The effect of increasing BW support from 100% BW to 20% BW on peak joint force reduction was more pronounced for the knee than for the ankle ($P = .007$ at 2.22 m/s and $P = .005$ at 3.3 m/s).

Thus, the estimated knee joint force at 20% BW corresponded to 23.8% (at 2.22 m/s) and 28.8% (at 3.33 m/s) of the initial values, while the estimated ankle joint force at 20% BW corresponded to 41.6% (at 2.22 m/s) and 52.8% (at 3.33 m/s) of the initial values at 100% BW (Figure 4).

In addition, step frequency at 2.22 m/s declined gradually from 163 steps/min at 100% BW to 126 steps/min at 20% BW. The corresponding values at 3.33 m/s were 168 steps/min to 135 steps/min ($P < .001$). The relative duration of the stance phase decreased from 74% of the running cycle to 43% of the running cycle at 2.22 m/s, and from 62% to 38% of the running cycle at 3.33 m/s ($P < .001$). Furthermore, increasing the running velocity from 2.22 m/s to 3.33 m/s increased cadence ($P = .005$) and decreased the relative duration of the stance phase at all levels of BW ($P < .001$).

Discussion

An integrated experimental approach was used to estimate neuromuscular and biomechanical effects of unweighting in a LBPP chamber. Running on the LBPP treadmill allows the biomechanical environment to be changed gradually in a well-controlled manner, and the results showed that participants adapted their muscle activation accordingly. The effect of unweighting on muscle activation was more pronounced for the knee extensor muscles than for the plantar flexor muscles. Hamstring activity (BF) remained unchanged. Unweighting reduced GRF_{peak} markedly. Model estimation of the effect of unweighting on peak joint force reduction, including contributions from external and internal sources, was more pronounced for the knee than for the ankle joint. Our data on running characteristic confirmed previous data.

Knee extensor and plantar flexor muscle activation decreased gradually when BW was increasingly supported, reflecting that

2A

2B

2C

2D

Figure 2 — Electromyography (EMG) from the vastus lateralis (VL), vastus medialis (VM), biceps femoris (BF), gastrocnemius (GAS), and the soleus (SOL) muscles. (A) EMG_{mean} at 2.22 m/s, (B) EMG_{mean} at 3.33 m/s, (C) EMG_{peak} at 2.22 m/s, (D) EMG_{peak} at 3.33 m/s. ($n = 8$).

activation of both muscle groups is dependent on the vertical forces acting on the participant during running. The relative decrease in muscle activity with increasing weight support was largely similar for the 2 knee extensor muscles. Likewise, the reduction in muscle activation was similar for the 2 plantar flexor muscles. However, the 2 muscle groups responded differently to BW reduction. Thus, the relative activation reduction of GAS and SOL was much less pronounced than the reduction of VL and VM activation, indicating different functions of the 2 muscle groups during running. The relative reduction of VL and VM activation corresponded largely to BW reduction, indicating that the main function of the knee extensor muscles are to resist gravity during running. In contrast, SOL and GAS activities were reduced much less than BW reduction. These data clearly show the dual function of the plantar flexor muscles as important contributors to propulsion and weight support against gravity. In contrast, muscle activation of the BF muscle remained unchanged despite increasing BW support as BF was mainly active during the leg swing phase, breaking the hip torque and as a stabilizer in the stance phase. Thus, using LBPP to manipulate the biomechanical environment systematically constitutes an

experimental exercise model that allows the specific function of the different muscles and joint-specific consequences for joint forces to be studied separately during running.

Previous studies have estimated joint compression forces between 8 to 14 times BW at the ankle and 7 to 11 times BW at the knee during running.^{7,8,10} The present ground reaction force was largely reduced in parallel with the increasing BW support as predicted in the hypothesis. These decreases contribute to a reduction of the lower extremity joint forces during running, although the internal sources, ie, muscle forces, are the major determinant of joint forces.¹⁰ Based on the present results on ground reaction force and EMG, peak joint force reduction was estimated, assuming near linear association between EMG, expressed as $\%EMG_{max}$, and muscle force for the specific muscle. The EMG/force relationship during dynamic contractions has been debated in the literature and should be used with caution as it depends on muscle length changes and contraction type.^{11,12} However, gradually increasing weight support while running on the LBPP treadmill does not induce changes in contraction type, muscle length, or major changes in muscle shortening/lengthening velocities. Therefore, it seems reasonable to

3A

4A

3B

4B

Figure 3 — Vertical ground reaction force as a function of body weight support ($n = 8$). (A) Mean ground reaction force (4–6 min). (B) Peak ground reaction force (4–6 min). Mean $\pm$ SEM.

Figure 4 — Estimated relative joint forces (knee and ankle) as a function of body weight support ($n = 8$). (A) 2.22 m/s. (B) 3.33 m/s. Mean $\pm$ SEM. ($n = 8$)

assume that the EMG/force relationship remains largely independent on weight support, which allows us to estimate relative changes in joint force using 100% BW as a reference.

Our EMG data indicated that the relative effect on the mechanical joint load of supporting BW in the LBPP chamber during running differed between the knee joint and the ankle joint. Thus, we estimated knee joint compression forces to be reduced largely in parallel with the body weight, as the GRF_{peak} was reduced to 28% (at 2.22 m/s) and 31% (at 3.33 m/s) of the initial values at 100% BW, and the EMG_{peak} of the VL and VM muscles was reduced to 22% and 28% at 80% weight support. In contrast, estimated relative ankle joint forces were less affected by weight support. Our model estimated a knee joint force of approximately 26% and an ankle joint force of approximately 47% at 80% weight support. Thus,

the study emphasizes that a change of the biomechanical environment through increasing weight support in the LBPP chamber has different effects on the knee and the ankle joint force due to the muscle-specific effect of BW support. These results should be taken into account in rehabilitation.

Noncontact anterior cruciate ligament injuries are most likely to happen during deceleration and acceleration of the body against gravity. The literature consistently shows that the anterior cruciate ligament is the major restraint against anterior shear forces applied to the tibia relative to the femur.^{13,14} Quadriceps muscle forces provide anterior shear forces on the tibia while cocontraction of the hamstrings is protective to the anterior cruciate ligament, as they provide a posterior shear force on the tibia.^{15,16} The potential role of the gastrocnemius muscle force for anterior cruciate ligament

strain is less clear.¹⁴ A change in the relative contribution of the knee muscle forces between the quadriceps, hamstrings, and gastrocnemius muscles is therefore expected to influence the net shear forces in the knee. The present results showed that the activity of the knee extensors is reduced significantly, while the BF muscle activity was maintained when unweighting the body. Thus, running with weight support changes knee muscle function toward facilitation of antagonist (hamstring) activity relative to agonist (quadriceps) activity. It is therefore expected that the anterior cruciate ligament tensile force due to the muscle contraction decreases significantly when running with BW support, which may protect the anterior cruciate ligament against injury.

The current study had inherent strengths and limitations. Estimates of joint loads are often based on data from a few subjects and single step cycles, and model limitations and assumptions regarding force distribution between muscles introduce nontrivial errors.^{7,8,10,17,18} The present experimental approach used to estimate relative joint force included data from internal (muscle activity) as well as external (ground reaction force) sources recorded across 3 minutes (378–504 steps) for each trial, which is a strength of the study. A limitation of this study is the assumption regarding the EMG–force relationship, which is complex during dynamic contractions as it depends on muscle length and contraction velocity and type.^{11,12} Another limitation is that only knee extensor and plantar flexor muscles were included in the model. The present effect of unweighting on joint forces is therefore discussed relative to the 100% BW condition only. The inverse dynamic method would be beneficial for determination of absolute joint forces. However, this is not possible with the LBPP device. Changing muscle coordination pattern can affect joint loads significantly even when the movement pattern is largely maintained. In the current study no indication of change in muscle coordination with increasing BW support was found in the medial-lateral direction (VL vs. VM) or when increasing running speed support. Hip muscles were not included in the current study, and it is therefore not known if the subjects changed the relative activation between knee extensor muscles and hip muscles with increasing BW support. This is a limitation of the study. Finally, joint angles were not measured.

For comparison with our results, we found 2 studies on the effect of body weight support on peak tibiofemoral forces, as provided by LBPP (Patil et al,³ $n = 4$; Macias et al,¹⁹ $n = 1$). The subjects were implanted with instrumented knee prostheses and net tibiofemoral contact forces were measured. Patil et al³ found that the peak tibial force at 25% BW relative to peak tibial force at 100% BW was 1.9 BW/5.1 BW, corresponding to 37% at 25% BW. Macias et al¹⁹ found an uneven load distribution between the instrumented leg and the healthy leg at high relative BWs and “high” velocity, tending toward more load on the healthy leg. This means that, in healthy subjects, with an even load distribution between left and right leg, we may expect values lower than 37%, as found by Patil et al.³ Macias et al¹⁹ found values of approximately 30% at 20% BW (based on readings from their Figure 6). Again, this value could be a bit overestimated due to the uneven weight distribution between the implanted knee and the healthy knee.¹⁹ For comparison, we estimated knee joint force at 20% BW to be 24% (2.22 m/s) and 29% (3.3 m/s) of the initial value at 100% BW for healthy subjects. Thus, our results regarding estimated joint forces mainly agree with previous results obtained with direct measurements of joint force in instrumented knee prostheses.

A number of training and rehabilitation modalities, eg, harness systems and swimming pools, are currently being used to

allow movements and exercise for patients with a need for weight support. However, the available systems are not optimal since harness systems may not be used over prolonged periods of time due to discomfort and impeded circulation. Movements in deep water may help to maintain the aerobic fitness, but the neuromuscular training and the movement pattern are very different from normal movements, as the forces applied to the body in water differ significantly from the forces during overground movements (eg,^{20–22}). Ambulation in a LBPP chamber may be beneficial not only for early mobilization after injury and surgery,²³ but also as a motor evaluation and learning tool for neuromuscular patients, as motor control can be challenged systematically.

In summary, running with BW support in a LBPP treadmill chamber allows the biomechanical environment and motor demand to be manipulated systematically. Participants adapted activation levels of the individual leg muscles to the new biomechanical conditions and maintained near normal movement patterns, while the estimated joint forces were reduced significantly. Estimated joint force reduction was more pronounced for the knee than for the ankle joint.

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