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A force measuring treadmill in clinical gait analysis

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Abstract
This preliminary study presents the development and testing of an instrumented treadmill device measuring the ground reaction forces (GRFs) and the feasibility of using this force measuring treadmill (FMT) in clinical gait analysis. A commercially available treadmill was modified and fitted out with three-dimensional strain-gauge force transducers. Tests of linearity, centre of pressure position (CoP), cross talk, natural frequency, background noises, and belt speed were undertaken in order to assess the performance of the FMT. In addition, the GRFs and segmental kinematics were recorded while healthy subjects and patients walked on the FMT, in order to compute the net ankle joint moments and the body centre of mass (CMb) kinematics and mechanics. The preliminary results of technical tests were satisfactory with an error less than 10% and dynamic tests in healthy subjects corresponded to the literature. The results of patients were clearly disturbed, demonstrating the ability of the FMT to discriminate pathological gaits from normal ones. We concluded that the GRFs measurements obtained from the FMT seem valid and the clinical assessment of net joint moments and CMb kinematics and mechanics seem feasible. The FMT could be useful device for clinical research and routine gait analysis since it allows gaining some extra room and quickly collecting the GRFs during a large number of successive gait cycles and over a wide range of steady-state gait speeds. However, more work is needed in this area in order to confirm the present results, collect reference data and validate the methodology across pathologies.

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Keywords: Ground reaction forces; Treadmill; Gait

1. Introduction
In clinical research and routine gait analysis, ground reaction forces (GRFs) are commonly recorded by means of floor-mounted force platforms (FMFPs). Raw GRFs data or computations based on GRFs, such as net joint moments and mechanics of the body centre of mass (CMb), are used to better understand the mechanisms of pathological gait [1–3].

With FMFPs, the assessment of kinetic data during several cycles is time consuming, since the patients must target their footsteps in order to allow adequate recordings, i.e. to put only one foot by individual plate. The assessment of the CMb mechanics also requires the use of large FMFPs or the combination of several FMFPs for measuring GRFs during several gait cycles. In addition to large financial means, this assessment implies a spacious gait laboratory.

One solution to overcome these drawbacks is to use a force measuring treadmill (FMT) since it allows gaining some extra room and quickly collecting GRFs during a large number of successive cycles and over a wide range of steady-state gait speeds [4]. Therefore, custom-made FMT devices were previously developed and various designs were proposed including mounting a force platform inside a treadmill [4,5], building a treadmill around a FMFP [6,7], and mounting a treadmill on top of multiple force transducers [8].

Until now these FMTs were principally used in fundamental research for monitoring GRFs during normal locomotion [8–10]. To our knowledge, few authors [11,12] used these treadmills for clinical research and none described the feasibility to use such a device for routine clinical gait analysis. Therefore, our purpose was three-fold: (1) to determine the FMT technical features; (2) to validate the kinematic and kinetic variables computed from FMT force in healthy subjects; and (3) to show the feasibility of using the FMT in clinical gait analysis.
2. Material and methods

2.1. Medical treadmill

A commercial treadmill (Mercury LT med, HP Cosmos, Germany) with an appropriate conveyor belt (1500 mm long × 500 mm wide) was used. The driving system provides a range from 0 to 22 km h\(^{-1}\) (by minimum increments of 0.1 km h\(^{-1}\)). The side handrails, the front rails, the user panel control, and the rear silent blocks supports were removed in order to significantly decrease the mass of the treadmill (190 kg) and keep the natural frequency as high as possible. Four customised 3D strain-gauge force transducers were designed to fit out the treadmill (Fig. 1A).

2.2. Technical tests

The FMT technical features were determined by tests of linearity, centre of pressure position (CoP), cross talk, natural frequency, background noises, and belt speed. All tests were repeated three times, and the vertical (\(F_v\)), forward (\(F_f\)), and lateral (\(F_l\)) forces were sampled at 100 Hz (500 Hz for the natural frequency). All technical tests were performed on raw GRFs.

2.2.1. Linearity and cross talk

The accuracy, linearity and cross talk for \(F_v\) were checked by successively applying four calibrated static loads (250, 251, 252, and 254 N) along a middle line and on the front, middle, and rear parts of the belt, up to a maximum of 1007 N. The non-linearity of the force outputs was assessed as the deviation (%) from the least-squares linear regression of the treadmill force outputs versus the applied loads.

The cross talk measured by applying a vertical force was determined along the longitudinal and lateral axes as the ratio (in %) between \((F_f \cdot F_v^{-1}) \times 100\) and \((F_l \cdot F_v^{-1}) \times 100\), respectively.

2.2.2. Centre of pressure position

The accuracy of the centre of pressure position was checked by applying a 501 N static load at different belt locations and quantified in forward and lateral directions as the maximum error (%) between the ‘real’ location of the load estimated by the coordinates of reflective markers filmed with a six-camera optoelectronic system (Gait El clinic 2000–2002, BTS, Italy) at 100 Hz and the location given by the force transducers. The location of the CoP in the forward (CoP\(_f\)) and lateral (CoP\(_l\)) directions were calculated as:

\[
\text{CoP}_f = \frac{d_1 \cdot F_{v\text{Rear}}}{\sum F_v} \quad \text{and} \quad \text{CoP}_l = \frac{d_2 \cdot F_{v\text{Left}}}{\sum F_v},
\]

where \(d_1\) and \(d_2\) are the distances between rear–front and left–right transducers, \(F_{v\text{Rear}}\) is the \(F_v\) from rear transducers, \(F_{v\text{Left}}\) is the \(F_v\) from left transducers, and \(\sum F_v\) is the sum of \(F_v\) from all transducers (Fig. 1A).

2.2.3. Natural frequency

The natural frequency was determined by tapping the centre of the belt with a mallet. A fast Fourier transform (FFT) analysis was applied to the force signals to determine the natural frequency of the FMT.

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Fig. 1. (A) Top view of the treadmill. The force transducers (1–4) are located at the four corners of the treadmill. (B) From top to bottom, ankle displacements, moments and power normalised by the subject weight during seven gait cycles in healthy subject #1 (thin line) walking on the FMT at 4 km h\(^{-1}\), and during four gait cycles in patient #1 (thick line) walking at 1 km h\(^{-1}\). Dashed lines indicate the zero. (C) From top to bottom, mean ± S.D. ankle displacements, normalised moments and normalised powers in healthy subject #1 (thin line, S.D.: grey areas) and patient #1 (thick line, S.D.: vertical bars) as a function of the normalised gait cycle (%).
Table 1
Characteristics of healthy subjects and patients

<table>
<thead>
<tr>
<th>Sex (M/F)</th>
<th>Age (years)</th>
<th>Weight (kg)</th>
<th>Height (m)</th>
<th>Computation (Mm/CM b )</th>
<th>Pathology</th>
</tr>
</thead>
<tbody>
<tr>
<td>Healthy subjects 1</td>
<td>F</td>
<td>23</td>
<td>65</td>
<td>1.71</td>
<td>Mm</td>
</tr>
<tr>
<td>2</td>
<td>M</td>
<td>21</td>
<td>75</td>
<td>1.70</td>
<td>CM b</td>
</tr>
<tr>
<td>3</td>
<td>M</td>
<td>26</td>
<td>85</td>
<td>1.80</td>
<td>CM b</td>
</tr>
<tr>
<td>4</td>
<td>M</td>
<td>29</td>
<td>95</td>
<td>1.75</td>
<td>CM b</td>
</tr>
<tr>
<td>Patients 1</td>
<td>M</td>
<td>57</td>
<td>85</td>
<td>1.78</td>
<td>Mm</td>
</tr>
<tr>
<td>2</td>
<td>M</td>
<td>58</td>
<td>76</td>
<td>1.74</td>
<td>CM b</td>
</tr>
<tr>
<td>3</td>
<td>F</td>
<td>21</td>
<td>57</td>
<td>1.65</td>
<td>CM b</td>
</tr>
<tr>
<td>4</td>
<td>F</td>
<td>63</td>
<td>74</td>
<td>1.75</td>
<td>CM b</td>
</tr>
</tbody>
</table>

F: female, M: male, Mm: net ankle joint moment, CM b : body centre of mass mechanics, R: right.

2.2.4. Background noises
The electrical noise was measured by comparing the treadmill force outputs with the motor turned off and on with the belt stationary. The mechanical and electrical noise caused by the belt speed irregularity and by the motor electric field was measured when the motor was turned on with the belt stationary and with the belt moving at different speeds (1–6 km h⁻¹). A FFT was applied to the force signals to determine the frequency content of the electrical and mechanical noises and to choose the appropriate filtering method for processing the force data (a digital moving-average filter, passed in both directions to effect zero-phase shift, was selected with a span of five points).

2.2.5. Belt speed
The belt speed accuracy was checked while an adult subject #4 (Table 1) walked on the FMT at 1–6 km h⁻¹. Several reflective markers glued to the edge of the belt were filmed by our optoelectronic system at 100 Hz. The forward velocity of marker was obtained by deriving the coordinates along the longitudinal axis. The maximum gait speed difference between the belt speed and the forward speed of the markers was calculated.

2.3. Dynamic tests in subjects
Dynamic tests were performed in four healthy subjects and in four patients. The GRFs and segmental kinematics were simultaneously recorded while the subjects walked on the FMT, in order to compute the net ankle joint moments for subject #1 in both groups (Table 1). All subjects were accustomed to treadmill gait before the walking tests. The gait of patients was characterised by a stiff ankle (patient #1-ankle osteoarthritis), a stiff-knee (patient #2-hemiplegia), a bilateral hip and knee flexions (patient #3-diplegia) and a decrease of the flexion-extension of the lower limb joints (patient #4-hydrocephalus).

2.3.1. Ankle muscular moment and power
The segmental kinematics was measured with the optoelectronic system at 100 Hz. Six infrared cameras measured the 3D coordinates of 19 reflective markers positioned on the specific anatomical landmarks [13]. The net sagittal ankle joint moment was computed by using a 3D inverse dynamics approach [13]. In order to estimate the vertical forces under each foot, a decomposition of superimposed vertical GRFs into left and right force profiles was computed by using the algorithm of Davis and Cavanagh [6].

2.3.2. CM b kinematics and mechanics
The 3D CM b kinematics was computed relative to the centres of mass of the individual segments by means of the optoelectronic system [1]. The methodology used to determine the CM b kinematics and mechanics has been described in detail in Cavagna [14]. In brief, the 3D accelerations of the CM b were calculated from \( F_v \), \( F_f \), and \( F_l \). The 3D velocities \( V_b \) were obtained by integrating the accelerations. The vertical (\( S_v \)), forward (\( S_f \)) and lateral (\( S_l \)) displacement of the CM b was obtained by integrating the velocities. The gravitational potential energy of CM b was computed from \( S_v \). The kinetic energies of the CM b were computed from the velocities. The total mechanical energy was computed as the sum of gravitational potential energy and kinetic energies. The external work (\( W_{ext} \)) during gait was determined by summing the increments of the total mechanical energy curve during a stride. The Recovery (\( R \)), quantifying the amount of energy-saving transfer between gravitational potential energy and kinetic energy of the CM b was also computed [15].

3. Results
3.1. Technical force treadmill tests
The results of the technical FMT tests are presented in Table 2. The maximum ‘non-linearity’ of force outputs was 2.1% in the forward direction. The maximum CoP position error was 2.8% (CoPf). The maximum cross talk resulting from a vertical load was 2.9% in the forward direction. The minimum natural frequency was 30 Hz in the vertical direction. With the belt moving at 3 km h⁻¹, the maximum background noise was 5.4 N. Finally, the maximum belt speed error was 7%.
Table 2
Results of the technical FMT tests

<table>
<thead>
<tr>
<th>Non-linearity (maximum %)</th>
<th>Vertical</th>
<th>&lt;0.01</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Forward</td>
<td>2.1</td>
</tr>
<tr>
<td></td>
<td>Lateral</td>
<td>1.1</td>
</tr>
<tr>
<td>CoP position (maximum error %)</td>
<td>CoP_l</td>
<td>2.8</td>
</tr>
<tr>
<td></td>
<td>CoP_f</td>
<td>0.3</td>
</tr>
<tr>
<td>Cross talk—vertical load (maximum %)</td>
<td>Forward</td>
<td>2.9</td>
</tr>
<tr>
<td></td>
<td>Lateral</td>
<td>1.3</td>
</tr>
<tr>
<td>Natural frequency (Hz)</td>
<td>Vertical</td>
<td>30</td>
</tr>
<tr>
<td></td>
<td>Forward</td>
<td>45</td>
</tr>
<tr>
<td></td>
<td>Lateral</td>
<td>35</td>
</tr>
<tr>
<td>Background noises (maximum N)</td>
<td>FMT turn off</td>
<td>Vertical</td>
</tr>
<tr>
<td></td>
<td>Forward</td>
<td>0.10</td>
</tr>
<tr>
<td></td>
<td>Lateral</td>
<td>0.15</td>
</tr>
<tr>
<td></td>
<td>FMT turn on</td>
<td>Vertical</td>
</tr>
<tr>
<td></td>
<td>Forward</td>
<td>0.19</td>
</tr>
<tr>
<td></td>
<td>Lateral</td>
<td>0.22</td>
</tr>
<tr>
<td>Moving belt (at 3 km h⁻¹)</td>
<td>Vertical</td>
<td>5.4</td>
</tr>
<tr>
<td></td>
<td>Forward</td>
<td>5.4</td>
</tr>
<tr>
<td></td>
<td>Lateral</td>
<td>2.4</td>
</tr>
<tr>
<td>Belt speed (maximum error %)</td>
<td></td>
<td>7</td>
</tr>
</tbody>
</table>

CoP: centre of pressure; CoP_l: CoP in forward direction; CoP_f: CoP in lateral direction.

3.2. Dynamic tests in subjects

The sagittal ankle displacements, the ankle muscular moment and ankle muscular power are presented in Fig. 1B for the healthy subject #1 walking at 4 km h⁻¹ and for the patient #1 walking at 1 km h⁻¹ during seven and four successive strides, respectively. The mean ± S.D. gait cycle normalised in percents are presented in Fig. 1C. The kinetic results obtained in the healthy subject are in agreement with the literature [1]. The results obtained in the patient clearly indicate a small ankle moment and lack of ankle power generation.

The mean ± S.D. (n=7) of CM_b phase portrait in the frontal plane between S_l and S_v is presented in Fig. 2A for the healthy subject #4 walking at 4 km h⁻¹ during seven successive strides. The CM_b displacement was simultaneously computed by means of optoelectronic system and the FMT. While the shape and amplitude of the CM_b kinematics in the frontal plane were close, the FMT measurements presented a smaller variation, especially in the vertical direction. The mean amplitude of S_v was 0.023 ± 0.004 m (optoelectronic system) versus 0.018 ± 0.002 m (FMT). The mean amplitude of S_l was 0.036 ± 0.006 m versus 0.032 ± 0.002 m. The coefficient of variation in the lateral and vertical directions was, respectively, 59% and 43% with the optoelectronic system versus 33% and 28% with the FMT. The CM_b kinematics computed by means of the FMT reduced the variation inter strides.

The CM_b mechanics for healthy subjects #2–4 are presented in Fig. 2B. The external work (W_ex) and the recovery (R) are presented as a function of speed. The CM_b mechanics results obtained in the healthy subjects are in agreement with literature (grey areas) [16]. The results obtained in the patients clearly show an increased external work accompanied by a worse energy transfer.

4. Discussion

All preliminary tests indicated that the quality of the force measurements seemed sufficient for appropriate interpretation of GRFs in pathological gait. The major difference between gait on a motorised treadmill or over ground gait is the frame of reference [17]. Previous studies [7,18,19] demonstrated that kinetic, kinematic, EMG, and metabolic variables of gait are almost identical in the two situations if the subjects are accustomed to treadmill walking.

The FMT provides many advantages over conventional FMFPs [7]. It allows a decrease in the data collection time and the space required, and to record GRFs at constant gait speeds. Moreover, it also allows recording the GRFs, the kinematics, EMG, and rate of oxygen consumption simultaneously.
Despite his numerous advantages, the FMT however has several disadvantages. First walking on a treadmill is more dangerous and some patients are anxious. Second the patients must be accustomed to treadmill walking before the assessment. Third the FMT records the summed GRFs from both feet. This is suited to measure the mechanical work done to move the CMb but not for the joint moments assessment. Nevertheless, the use of a simple algorithm [6] allows separating the individual foot vertical forces before computing joint moments. Another solution is to use a FMT with split belts [20] and force transducers. However, this design compels the subject to walk with a wider base of support.

Clinicians are sometimes suspicious of gait analyses results collected on a FMT, arguing that treadmill gait is different from over ground gait. In routine pre and postoperative gait assessments and follow-ups, it is easy to avoid the critics by realising all the assessments on the FMT. Therefore, the usefulness the FMT device should be judged not solely on its own disadvantages but on the balance of its advantages and disadvantages. While the present study demonstrates the feasibility of using the FMT in clinical gait analysis, more work is needed in this area in order to confirm the present results, collect reference data and validate the methodology across pathologies.

Acknowledgements

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