UNIVERSITY OF BELGRADE
FACULTY OF ELECTRICAL ENGINEERING

Motor drive for control of pelvis motion during gait

Diploma work

Neural Prostheses

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Acknowledgments

This work was done in Laboratory for Biomedical Engineering at Faculty of Electrical Engineering. This is part of project Walkaround®, development of the walking aid for rehabilitation, in cooperation with Faculty of Mechanical Engineering.

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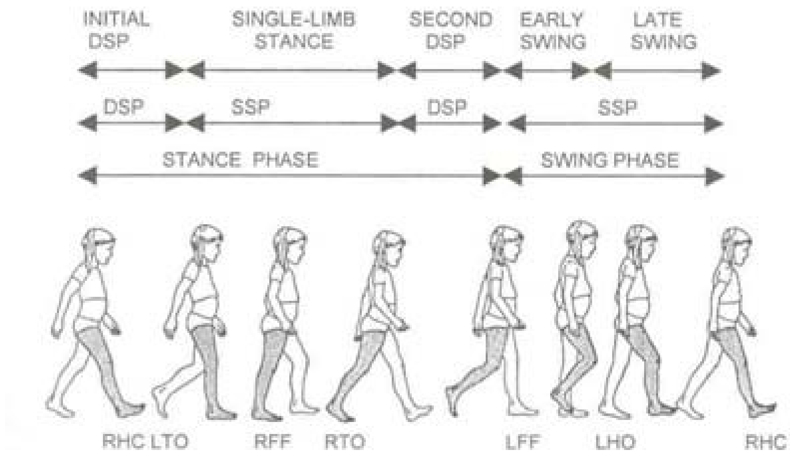
Goal

Goal of this work is to evaluate possible solution for motor drive, which controls pelvis motion during gait. Motor drive is under development for use in rehabilitation apparatus Walkaround®. Walkaround® is mechanical walking aid designed for gait revival after stroke or other disability. Instead passive support now, with motor drive for control of pelvis motion during gait, Walkaround® will have active gait training ability. This way it could be simulated normal pelvis movement during gait of healthy person which is very important to obtain normal gait parameters during rehabilitation after injury.

Introduction

Human displacements have been widely studied and described in the literature since the 19th century. Various invariants have been extracted from walking structure. In steady state, walking is symmetric and periodic. The walking cycle can be divided into strides, which are themselves, divided into steps. Each cycle is composed of single and double supports. The walking structure can vary from one individual to another depending on their sex, age, weight and height. A same individual also presents various walking patterns depending on learning and tiredness, which are a source of modification of the gait. The displacement parameters also vary with the environment characteristics such as obstacles, stairs, narrow supports (beams), etc. Transient walking corresponds to short modifications of a given pattern: starting and ending steps, turning, load carrying, etc. The objective of the movement also influences the gait; to efficiently cover a long or a short distance, the walking parameters such as step length and speed are different.

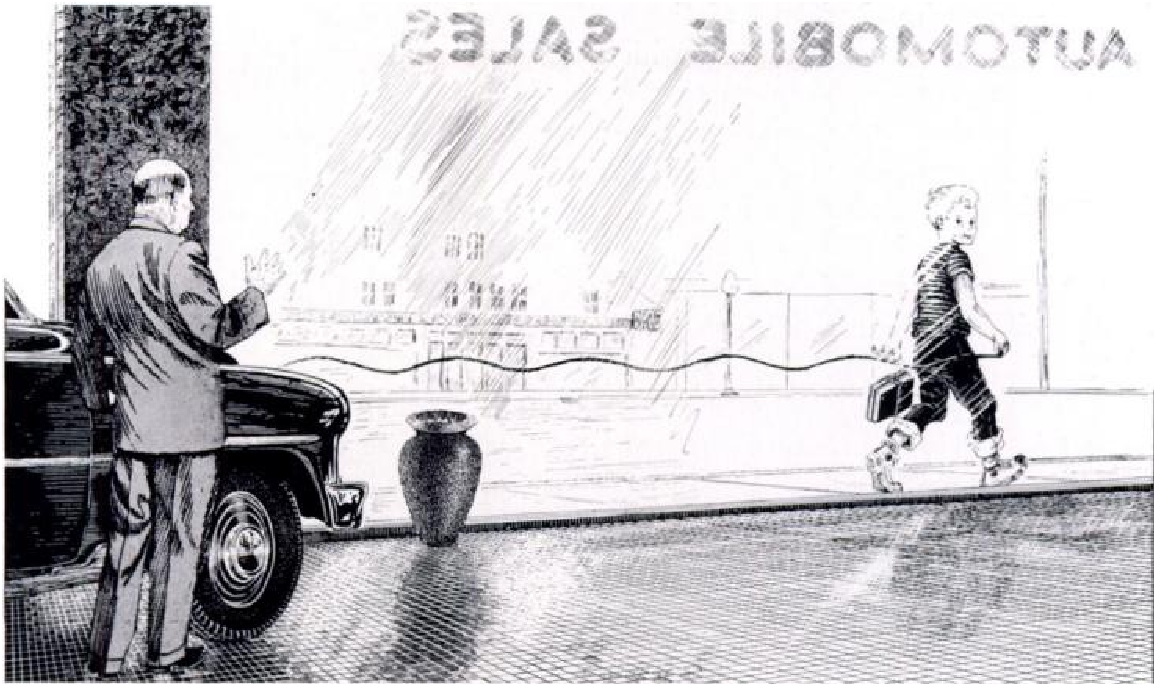


Picture 1. Gait phases

The successive stages involved in the walking control process can be summarized as follows: generation of a displacement idea at the central nervous system (CNS) level; transmission of the signals corresponding to walking to the muscles via the spinal cord; contraction of the muscles involved in locomotion; regulation of the forces and joint torques by the central nervous system on the basis of sensor information. The efficiency of this chain highly depends on the perceptive abilities of the human system.

In order to present a simplified concept of the phenomena of locomotion, it is convenient to consider the behavior of the center of gravity of the body during the cycle of motion. By so doing an abstract conception is introduced, since it is assumed that the entire weight of the body is concentrated at that point.

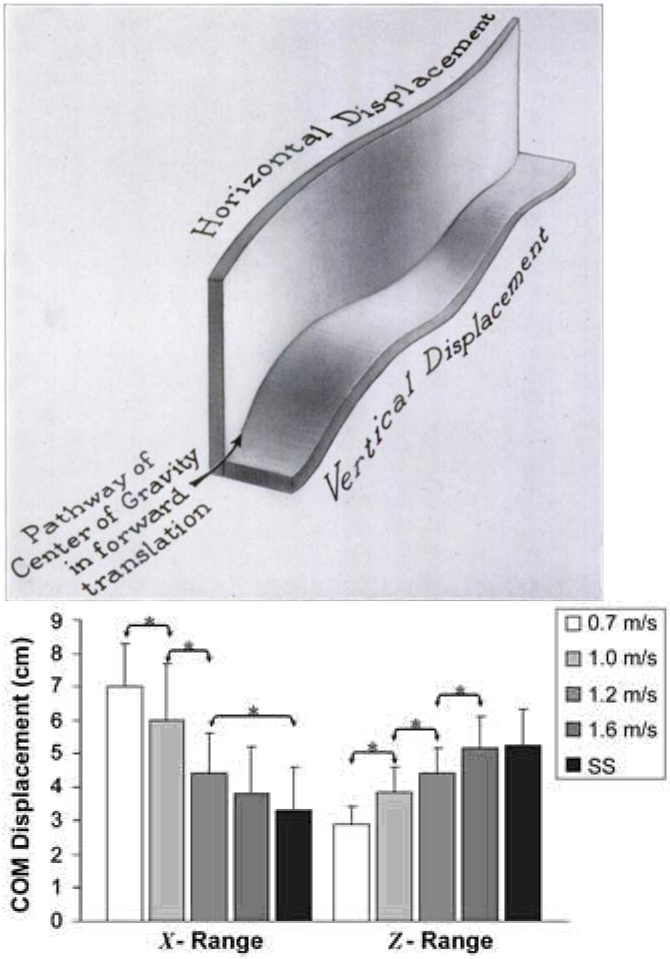
During human walking, the center of mass (COM) translates along the direction of travel but also moves in a sinusoidal pattern in the vertical and lateral directions. In both the vertical and lateral directions, two maxima appear: the first near 30 percent of the gait cycle in single-limb stance and another near 80 percent of the gait cycle in mid-swing; minima appear at 0 percent and 100 percent of the gait cycle in loading. Therefore, the COM reaches its highest and most lateral point as it passes over the planted foot and its lowest and most central point passing from one foot to the other. Since human walking is staggeringly complex, analysis of the COM movements has been suggested to simplify and illuminate the disturbances due to a broad range of pathologies.



Picture 2. The line drawn on the glass window represents the approximate pathway of the center of gravity of the body

In normal level walking, the center of gravity of the body describes a smooth regular sinusoidal curve in the plane of progression. From this curve, it is determined that the center of gravity is displaced twice in a vertical direction during the cycle of motion from the position of the heel-strike of one foot to the subsequent heel-strike of the same foot, that is, as the body passes successively in a double pace over first the right limb and then the left. The total amount of this vertical displacement in normal adult males is about four and half centimeters; individual variations are so small that they may be neglected. The summits of these oscillations occur at 25 and 75 percent of the cycle, each corresponding to the middle of the stance phase of the supporting limb; the opposite limb is at this time at the middle of the swing phase. At 50 percent, or the middle of the cycle, the center of gravity falls to its lowest level; this position corresponds to the interval of double weight-bearing when both feet are in contact with the ground. As the curve is followed in its ascent and descent, it is found to fluctuate evenly between the maximum and minimum of the displacement, with few if any irregularities.

The center of gravity of the body is also displaced laterally in the horizontal plane. Relative to the plane of progression, the center of gravity describes a sinusoidal curve, the summits of which alternately pass to the right and to the left in association with the support of the weight - bearing extremity**.** The curve is smoothly undulating without irregularities and it is similar in form to that of the vertical displacement.



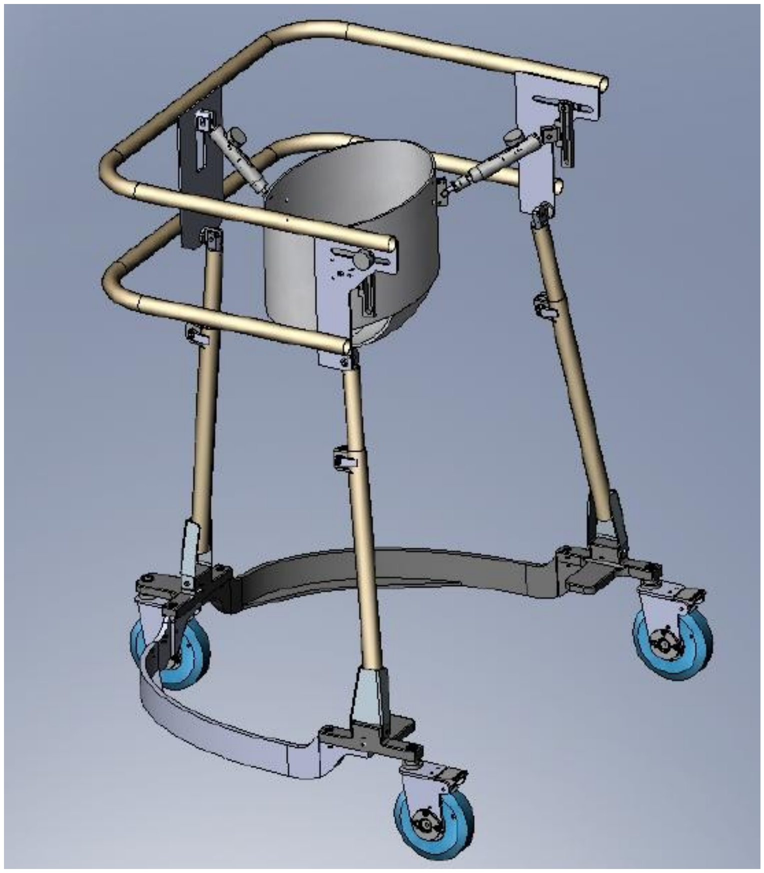
Picture 3. The intersection of the horizontal and the vertical displacements produces the pathway of the center of gravity in locomotion and its values for different walking speeds (center of mass (COM) vertical (*Z*) and lateral (*X*) displacement, SS – self selected)

Among adults, stroke is the third leading cause of death and the most common cause of disability. The recovery of function after stroke has been well studied. For example, most recovery of physical function usually occurs during the first 3 months after stroke, with the majority of the gains during this 3-month period often occurring within the first 30 days. Regaining the ability to walk is a major goal during the rehabilitation of stroke patients. Factors important to reach this goal are early, functional, goal oriented intensive training. In clinical practice direct assistance during exercise is intentionally restricted and strong personal involvement of the patient is encouraged. Locomotor training studies have shown that the human lumbosacral spinal cord remains capable of responding to sensory information related to locomotion even when damaged. This phenomenon is assigned to circuits within the spinal cord that act as central pattern generators. These circuits exhibit neural plasticity and are capable of motor learning if properly activated. It is maintained that for activation to occur, specific sensory input associated with performance of a motor task must be provided and followed by repetitive practice of the task. These requirements are fulfilled during locomotor training.

Walking aids are often used to maintain safety and increase independence during gait training.

Method

At Faculty of Electrical Engineering apparatus for locomotor training Walkaround® is under construction. It is used for gait rehabilitation of stroke patients. For faster recovery it is very important to exercise gait in reproducible patterns. This way central pattern generator will adopt gait parameters for muscle activation. Manual training is lacking in repeatability, therefore it is not an optimal way of performing gait rehabilitation. The job of a physical therapist is a demanding one. Clinicians use their own bodies to lift, move, and provide “safety nets” for patients who may be three times larger than the clinicians. The intensity and duration of physical therapy sessions is often limited due to simple exhaustion of the clinician. Safety concerns prevent the clinicians from challenging the patient as much as they could to enhance learning since falls or other injuries are not acceptable. Response to these problems is achieved using walk aids.



Picture 4. Walkaround®

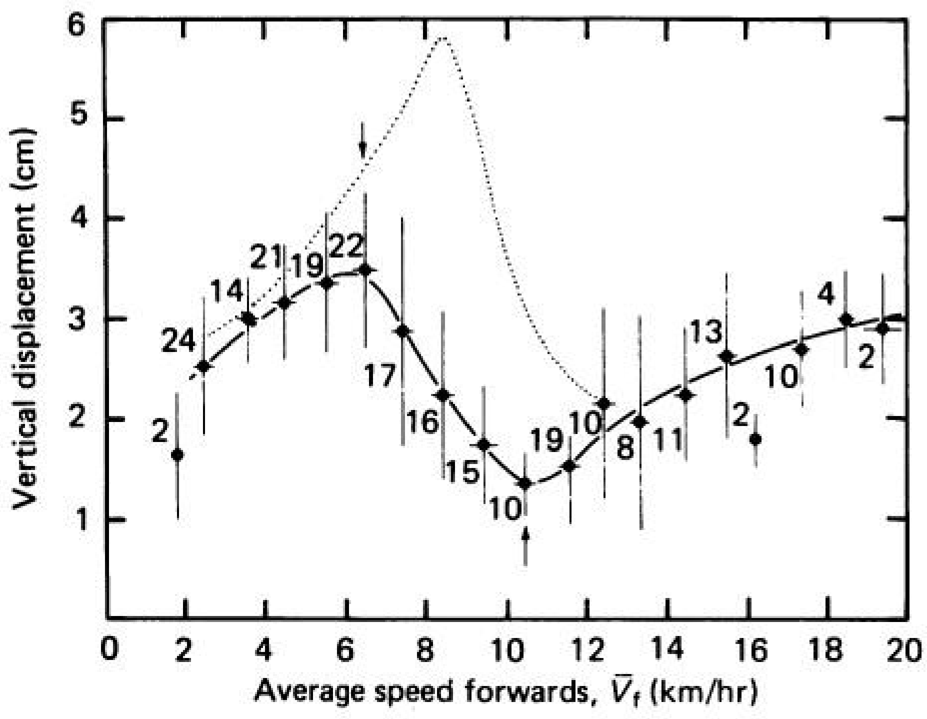
The Walkaround® was designed for facilitating of the exercise of walking of large population of individuals with post-stroke hemiplegia in acute or sub-acute phase; yet, it could be used for other individuals with disability who need postural assistance.

Task-oriented exercise was shown to contribute to the recovery of function in individuals with impairment due to neurological or orthopedic problems. It was demonstrated that weight-supported treadmill training improves walking in individuals with locomotor disorders. Manually assisted treadmill training has been introduced for individuals after spinal cord injury (SCI) and cerebro-vascular accident (CVA). The reports of the implementation of this therapeutic modality are somewhat controversial, especially in individuals after SCI. In individuals with hemiplegia weight-supported treadmill training suggests better balance, lower-limb muscle recovery, increased walking speed after the exercise, prolonged ability to walk, as well as symmetry and stride length. In studies with a limited number of patients, positive therapeutic effects were reported in individuals with multiple sclerosis (MS), Parkinson's disease, and traumatic brain injury (TBI).

It is suggested that partial weight-supported treadmill training in severely impaired post-CVA individuals led to better walking and postural control compared with individuals who were trained without the body support. In systematic reviews the efficacy of body-weight supported treadmill training for post-CVA individuals indicated conflicting evidence among the different studies. Three meta-studies suggested that only after adequately sized randomized clinical trials could final conclusions fully justify this method of supported walking. It was concluded that the use of a treadmill provides individuals with prolonged and safe training; thereby, possibly attaining better outcomes.

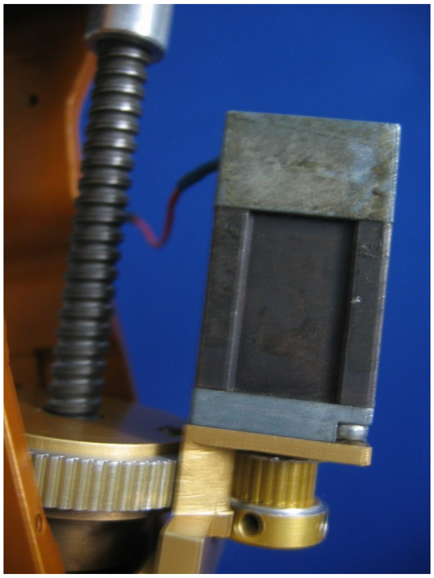
Walkaround® is compliant walker for humans having limited use of their legs and lower back includes an upright wheeled frame which at least partially surrounded an upright user wearing a partial body harness which is attached to the frame by means of cable compliant apparatus consisting of sets of cable segments and angle bracket members connected between opposite side members of the frame and adjacent side portions of the harness. The frame itself lends itself to several embodiments, one of which completely surrounds the user, while the other consists of a frame open at the front and including a pair of upright side members which attach to the cable support apparatus and which is generally adjustable to accommodate the user's height. Three suspensors form the suspension system. The suspension system was designed in form of a four bar mechanisms to control not only the force, but also the torques between the rigid frame and ergonomic lumbar belt.

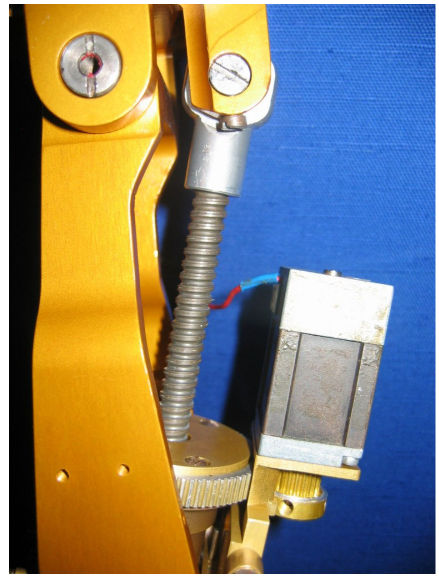
Improvement of this system is adding active control of pelvis motion during gait. It is necessary for achieving normal gait movement. This way, central pattern generators will adopt functional gait pattern faster. Movement of the pelvis can be modeled with center of mass displacement. This is very simple model but it can satisfactorily demonstrate gait parameters. Displacement of the center of mass of the body is affected by walking speed.



Picture 5. The vertical displacement of the center of gravity during each step is plotted as a function of walking speed. Each point represents the average of the data obtained from the eight subjects studied; the bars indicate the standard deviation of the mean, and the figures near each point give the number of items in the mean. The dotted line gives the trend of the data obtained in normal walking by Cavagna et al. (1976).

For normal walking speed (about 1m/s) vertical displacement of the center of mass is around 3 cm. This data will be used for control of pelvis motion during gait. So it is necessary to move center of mass in reproducible manner, at natural amplitude during every step. In Walkaround® system it is achieved by shortening front bar with motor drive and a ball-screw mechanism.





Picture 6. Motor drive with ball-screw mechanism

In Walkaround® geometry it was requisite to calculate front bar range of movement for reciprocation of the center of mass. It is assumed that the center of mass is located in center of pelvis, so moving of the center of mass corresponds to moving of pelvis. Because all of suspensors that are holding harness are the same, the center of mass will be located in center of triangle made of attaching places of suspensors. In order to find necessary shortening of the front bar to obtain functional center of mass movement I had to solve system of analytic equations. System of equations is made from geometrical parameters of the Walkaround®. System of equations is solved parametrically, so it depends on subject’s height and desired elevation of pelvis.

System solving:

$$\mathrm{Solve}\left[\left\{d == A - p,\ d == \sqrt{(A-q)^2 + (H-B)^2},\ H - B == 3e,\ r == \sqrt{p^2 + H^2} - \sqrt{q^2 + B^2}\right\},\ \{r\},\ \{q, B, p\}\right]$$

r - shortening of front bar
e - movement of pelvis
d - dimension of metal holding frame
H - patient’s pelvis height
A, B, p, q - geometrical points

This system of equations has 8 solutions, but only one satisfies physical constraints:

$$-\sqrt{\Big(2A^2-2Ad+2d^2+2A\sqrt{d^2-9e^2}-6eH+2H^2-2\sqrt{\Big(A^4-2A^3d+2A^2d^2-2Ad^3+d^4+\frac{2A^3d^2}{\sqrt{d^2-9e^2}}-\frac{4A^2d^3}{\sqrt{d^2-9e^2}}+\frac{2Ad^4}{\sqrt{d^2-9e^2}}-\frac{18A^3e^2}{\sqrt{d^2-9e^2}}+\frac{36A^2de^2}{\sqrt{d^2-9e^2}}-\frac{18Ad^2e^2}{\sqrt{d^2-9e^2}}-6A^2eH+12AdeH-6d^2eH+2A^2H^2-2AdH^2+2d^2H^2+\frac{2Ad^2H^2}{\sqrt{d^2-9e^2}}-\frac{18Ae^2H^2}{\sqrt{d^2-9e^2}}-6eH^3+H^4\Big)}\Big)}$$

After simplification solution is in form:

$$-\sqrt{2}\sqrt{\Big(A^2+d^2+A\left(-d+\sqrt{d^2-9e^2}\right)-3eH+H^2-\sqrt{(A^2-2Ad+d^2+H^2)\left(A^2+d^2+2A\sqrt{d^2-9e^2}-6eH+H^2\right)}\Big)}$$

From geometry parameter A can be represented as:

$$A=W-\sin\alpha\cdot H$$

where: W is width of frame base and α is angle of front bar (7^0).

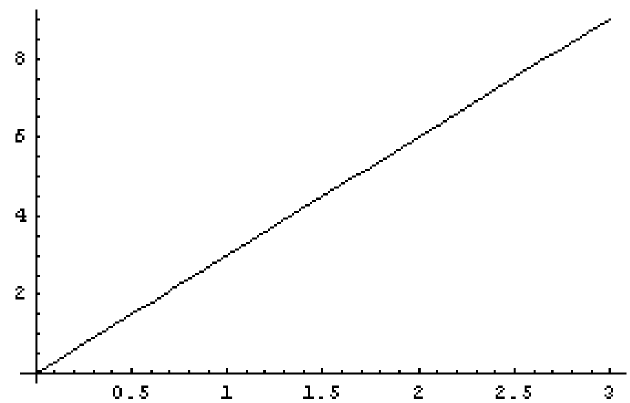
Now geometrical parameters of patient can be applied. For parameters:

$$H=70$$
$$d=55.6$$
$$A=60-0.12\cdot H$$

solution depends only on e.

$$\sqrt{\Big(15557.5-420e-102.938\sqrt{3091.36-9e^2}-140.244\sqrt{10640.4-420e-102.938\sqrt{3091.36-9e^2}}\Big)}$$

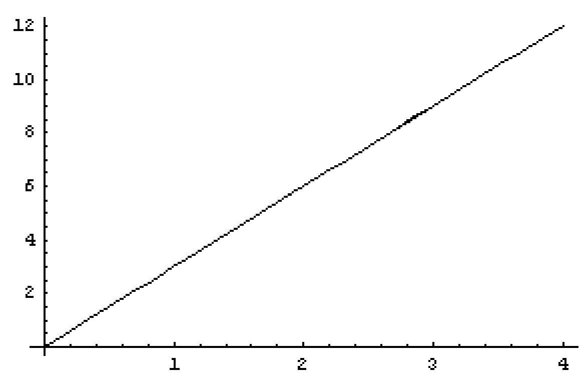
This solution is still too complicated and it is hard to see nature of dependence between shortening of front bar and movement of the center of mass. Next step is to draw function r(e).



Picture 7. Displacement of center of mass by shortening of front bar

It looks linear so I tried to fit linear approximation into solution. After fitting function solution was in form of:

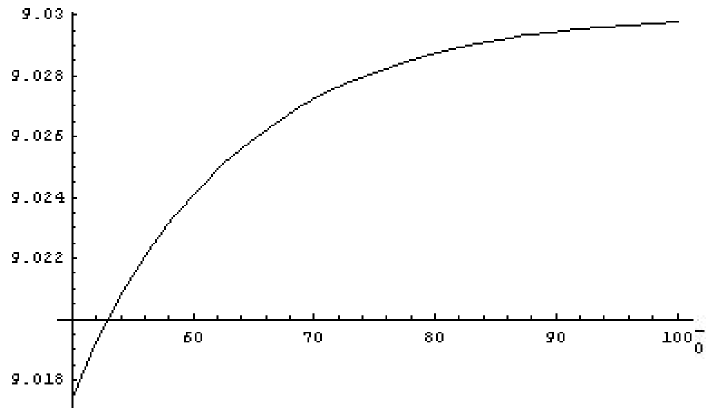
$$\boxed{r = 3.013 \cdot e}$$



Picture 8. Overlaying original function and fitted function

Maximum difference between original solution and fitted function on preferred interval for given parameters is 0.018 cm.

Now it will be calculated effect of patient's height on determining necessary front bar movement. For patient's pelvis height from 50 cm to 100 cm is plotted shortening of front bar to reach 3 cm movement of the center of mass.

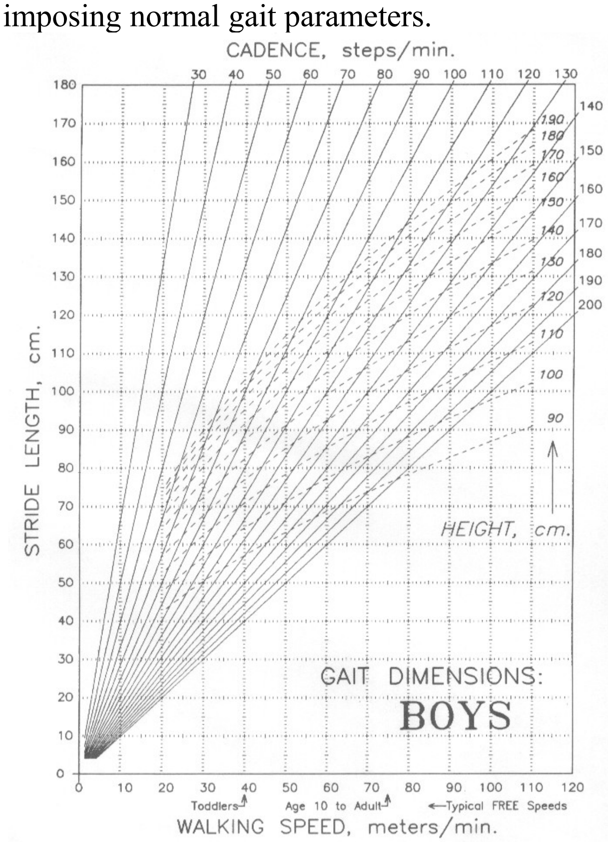


Picture 9. Change of shortening of front bar to reach 3 cm center of mass displacement for pelvis height from 50 cm to 100 cm

Maximal difference between necessary front bar movement for patient's pelvis height from 50 cm to 100 cm is about 0.012 cm.

This small difference could be neglected for practical purpose.

Next thing that needs to be calculated is period of step for different walking speeds. During normal gait step length and step period is dependent on walking speed. Measure of degree of recovery is attained walking speed. Faster walking speeds indicate that gait function is restoring. For achieving fast gait recovery after stroke, it is essential to train locomotion system by imposing normal gait parameters.



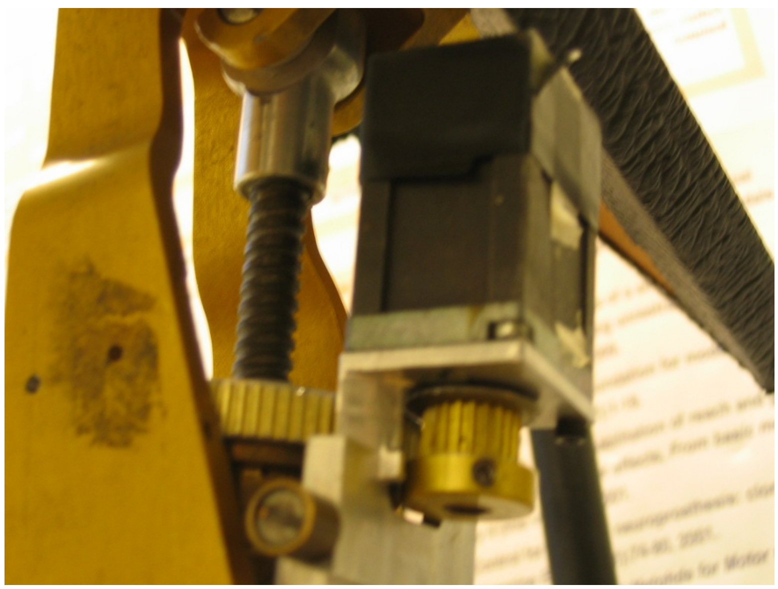
Picture 10. Gait parameters

By extrapolating data for normal walk on slow walking speed step cycle duration will increase to about 1.8 s.

Now, task is to move center of mass in appropriate time for different walking speeds.

The motor block provides helical motion with each step producing a 2.8 mm step. The DC motor peak is Pmax = 96 W, the nominal power is P0 = 38 W, the maximal torque is Mmax = 75 mNm, the maximum speed nmax = 5000 rpm , the back EMF is 8.2 mV/(rad/s), and the transition time from zero to nmax is tr = 16 ms.

The rigidity control block is implemented as a drum brake with asbestos lining. The friction coefficient asbestos to metal is μ = 0.54. A DC micro-motor drives the mechanism with the peak torque of Mmax = 13 mNm; its efficiency being 74%, the maximum speed nmax = 9500 rpm. The motor is directly coupled to the gear box with a reduction ratio of n = 97.3. The maximum moment upon the axle of the reduction gear is M = 0.9 kNm. The maximum force exerted on the lining fixed end, caused by the friction effect is 1.18 kN. The transition cycle is t = 12 ms. The mass of the actuator without the power supply is 456 g. The special features of the DC motor include long-life commutators and the provision of some additional damping during ballistic movement.



Picture 11. The motion motor with gears

The motion motor is responsible for the movement of the front bar to simulate natural gait at its maximum capacity (without load) with a 12 V supply runs at 4721 rpm. This motor is directly connected to a pinion gear A which transfers the speed to a sprocket gear B by means of a belt.

Characteristics	Gear A (pinion)	Gear B (sprocket)
Outer diameter	5.125 cm	5.125 cm
Tooth width	0.73 cm	0.70 cm
Size of teeth	-	0.1 cm
Number of teeth	22	80

The relationship R that exists between the two gears mentioned is given by:

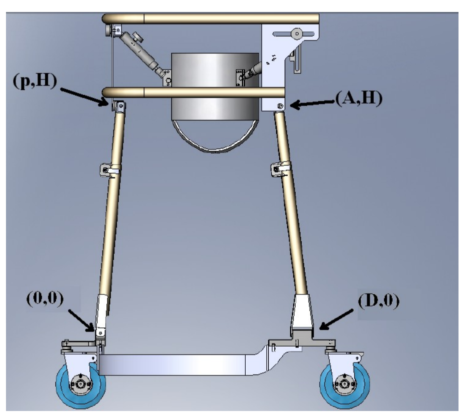
R =80/22

Where R = Number of teeth in gear A / Number of teeth in gear B

R = 3.6363

A (N. Turns)	B (N. Turns)	Ball Screw (cm)
0	0	0
4.20	1.15	0.43
8.42	2.32	0.85
12.65	3.47	1.30
16.85	4.63	1.70
21.05	5.80	2.15
25.27	6.95	2.57
29.48	8.10	3

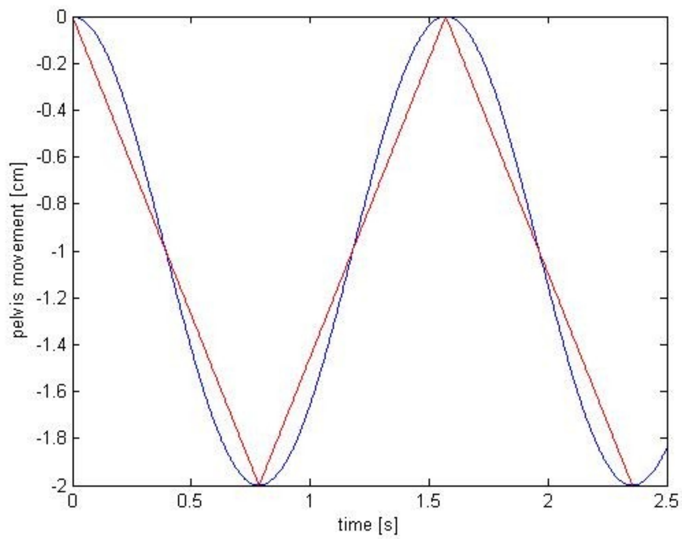
Although pelvis during gait is moving sinusoidal, in this work is moving linear from maximum to minimum position. For calculated shortening of front bar of about 9 cm motor will make 88.44 turns. Depending on walking speed motor will turn in different speeds. For characteristic walking speed of 0.3 m/s it will be determined motor rpm.



Picture 12. Walkaround®

For walking speed of 0.3 m/s step duration is about 1.8 s, so pelvis movement will last for 0.9 s. During this time pelvis will reach maximal displacement. For maximal displacement of 3 cm motor needs to go on 5896 rpm. This is more than motor can achieve. But for slower speeds displacement of center of mass is getting smaller. Instead of 3 cm in this purpose can be used 2 cm for vertical displacement of pelvis. Now, shortening of front bar is 6 cm and motor will make about 59 turns. For this parameter motor needs to go on 3931 rpm. This value is achievable.

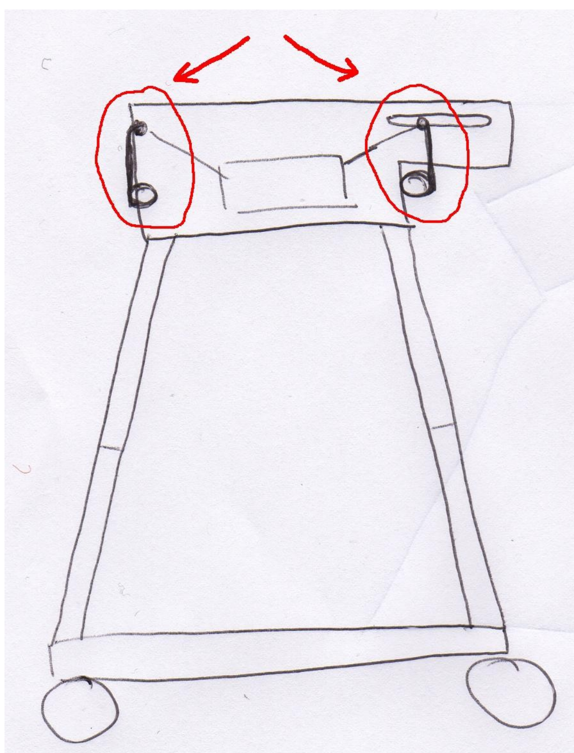
Greater speed of walking and normal speed cannot be reached with this construction while sustaining natural gait parameters.



Picture 13. Natural (sinusoidal) pelvis movement (blue) and imposed pelvis movement (red)

Alternatives

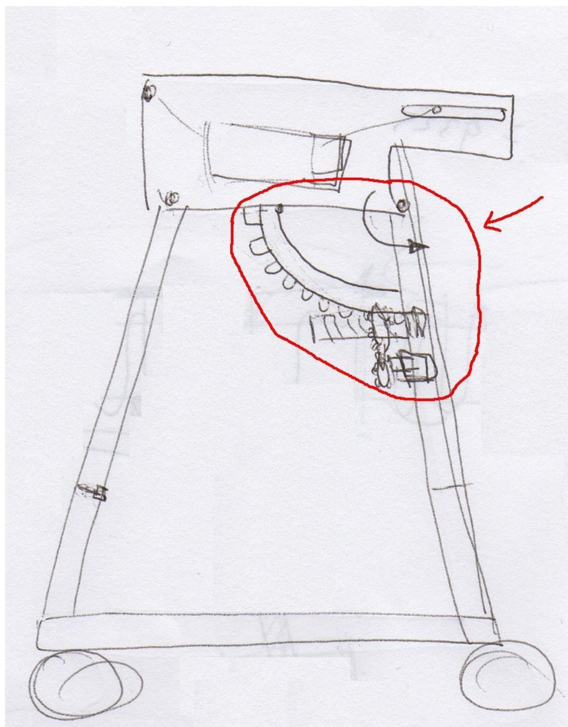
As system described above has deficiency in achieving normal speed of walking, it is necessary to devise new system that can overcome this demerit. This new system could work with original system or be stand-alone.



Picture 13. Alternative system 1

Original system controlled shortening of the front bar to achieve pelvis movement.

The first alternative system should be orientated in controlling harness movement. It is attached to the all three suspensors by cable. By loosening the cable pelvis will move in desired direction. For development of this system it is necessary to make actuators system and mechanism of control. With this improvement it would be possible to make new modes of training like Challenge (allows the clinician to challenge the patients balance as he is walking by loosening or widening the zone of safety such that the patient can explore the limits of his/her stability while maintaining safety) or Perturbation (while the patient is performing either static or dynamic activity the device allows for the clinician to throw the patient off balance).



Picture 14. Alternative system 2

The second alternative system is not based on front bar shortening than on tilting the frame. Upper part of the frame will change its relative angle to back bars so it will produce tilting of the harness and lowering pelvis. For development of this system it is necessary to make actuators system and mechanism of control.

Discussion

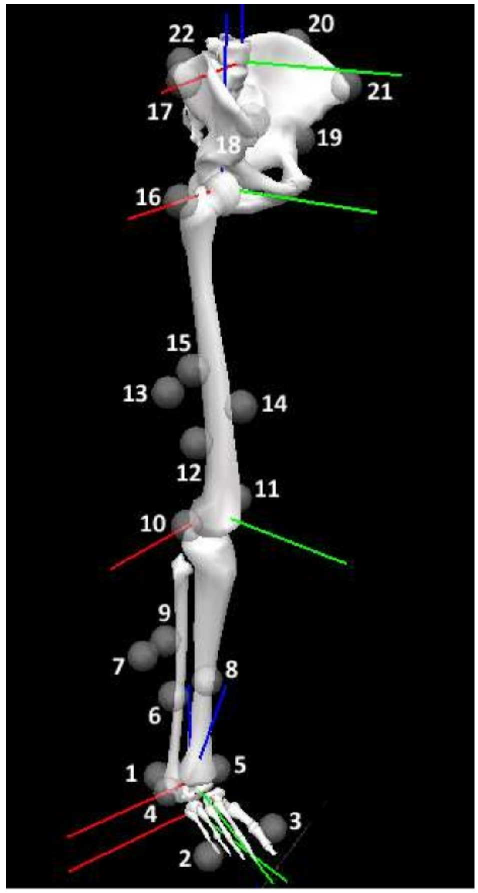
In this work was described system for motor drive for control of pelvis motion during gait. System is currently under development for use in walking aid Walkaround®. Walkaround® is now passive walking support device, but for better rehabilitation results it should have active gait control. This, active, system is designed to simulate natural pelvis movement for exerciseing gait in reproducible patterns. This is done because central pattern generator will adopt gait parameters for muscle activation faster, which leads to faster recovery.

Althou this system can achieve walking speeds of about 0.3 m/s, it is still enough for regular rehabilitation purposes. It reaches its full potential for slow walking speeds, but this speeds are characteristical for start of rehabilitation. Most important time for recovery is that time, just after stroke. This is hardest time for the clinicians and walking aid of this type is very desirable.

Appendix

Gait Data Collection

The acquisition system in this case was selected to be motion camera based, in a laboratory setting. The gait laboratory which was used for acquisition is located at the Center for Sensory-Motor Interaction, Aalborg University, Denmark. It consists of eight Qualisys ProReflex™ cameras (www.qualisys.com) used for motion capture, one AMTI force platform (www.amti.biz) for ground force measurement, Qualisys Track Manager (QTM) and C-Motion software for acquisition. The motion camera system operates by tracking reflective markers placed on a test subject's anatomical landmarks on the foot, leg segments, and pelvis. Marker placement was completed as per the manufacturer's instructions; 22 markers were used in total. Note that some error may exist in their positioning due to anatomical variations in individuals, but it should not be significant since the placements were attempted to be made as precise as possible.



Reflective Marker Placements. The reflective markers places on the right leg of the test subject are shown.

Notice that the eight cameras are all focused on the centre of the force platform. Since the force platform can provide GRF data for only one step in each acquisition run, it is important that the cameras capture motion most accurately around that area. It can provide GRF data in both the horizontal and vertical direction as required by the simulation, as well as the amount of force exerted upon it by the individual during stepping. These values are all important for computing the elastic conformity and behavior defined for the foot.

The way in which the camera and force platform data is converted into usable data is through the QTM software. The software records the data from the combined sensor inputs and develops a dynamic trajectory profile of the system based on the markers placed on the subject. The profile is based on the motion of the markers as captured by the cameras. This profile can include a substantial amount of raw data error making it is necessary to manually examine the marker trajectories in order to verify and minimize the profile errors.

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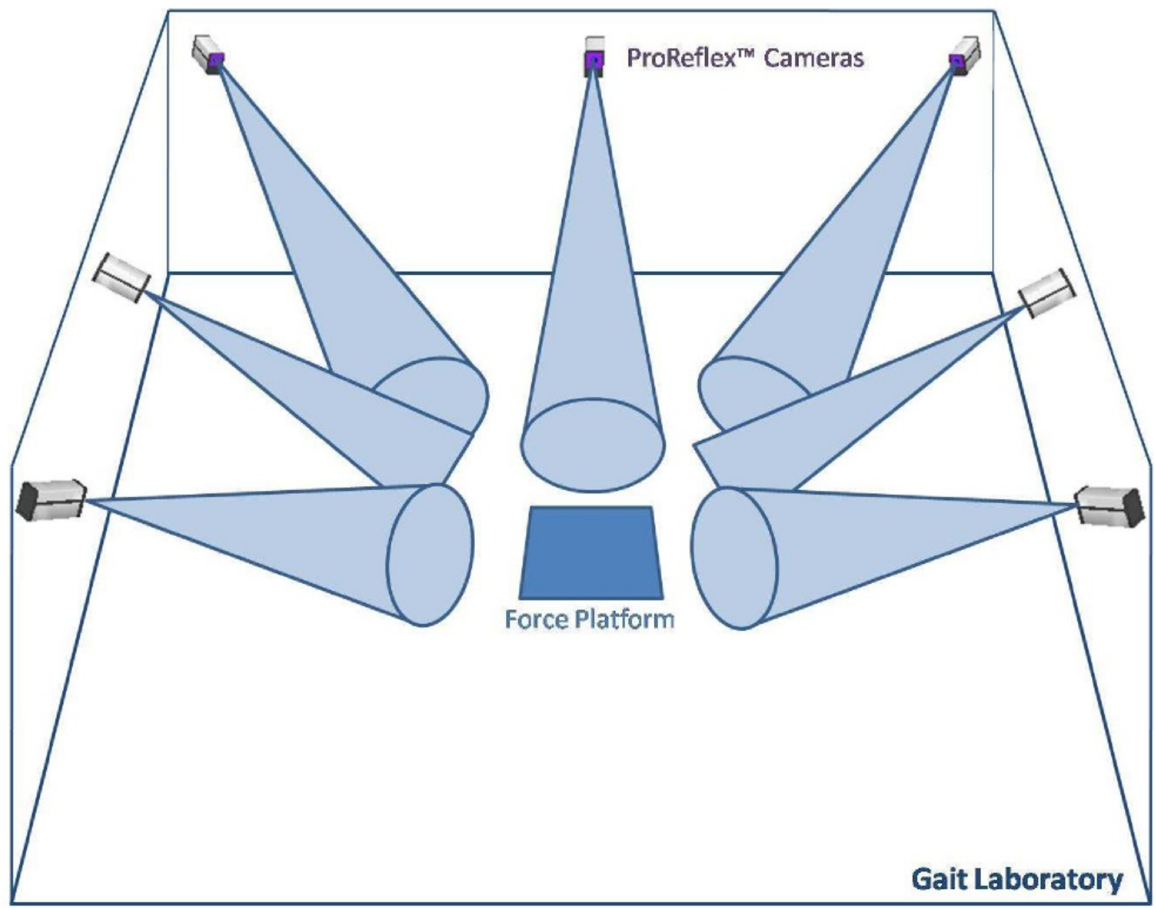
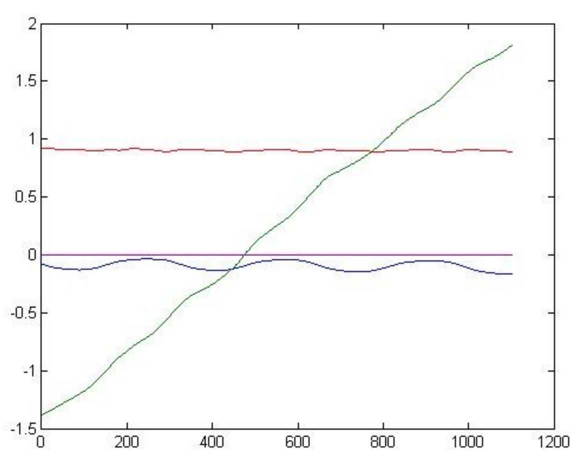
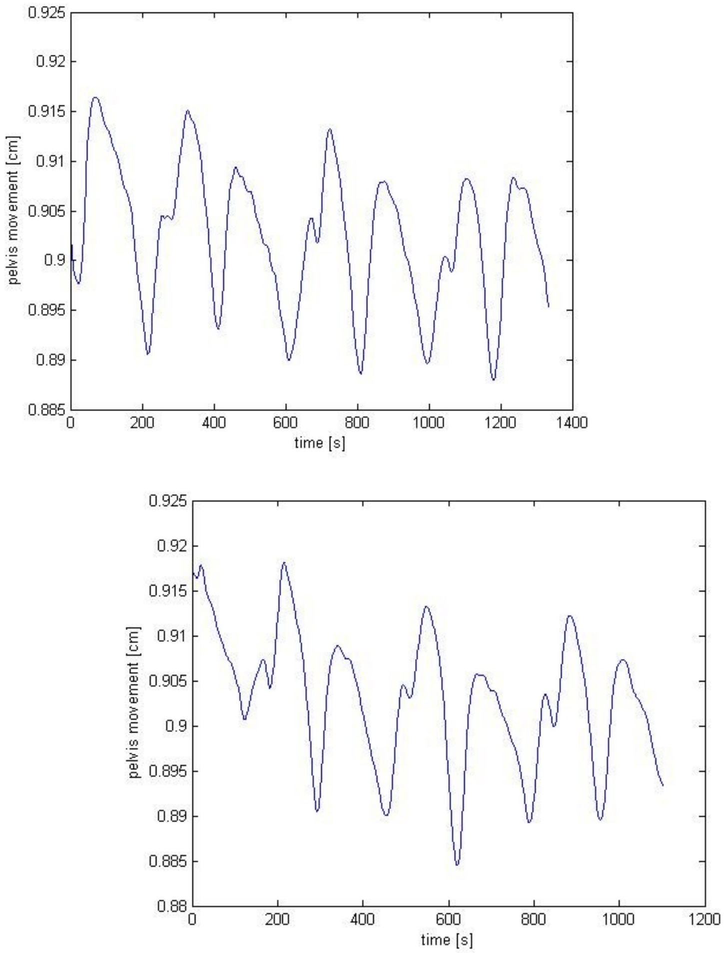


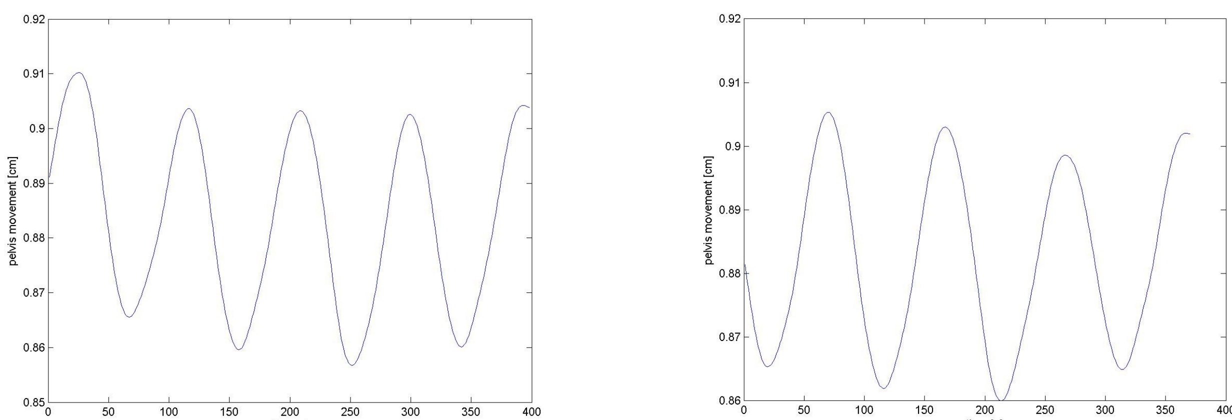
Gait Laboratory Environment. The diagram shows 7 ProReflex cameras focusing in the general direction of the lone force platform. The area which the cameras capture is later calibrated for precision. Note that there are in fact 8 cameras in the actual lab, but one was omitted to prevent image clutter. This image is not to scale.



Hip movement for slow walk



Pelvis movement in vertical direction for slow walk



Pelvis movement in vertical direction for normal walk

As seen on previous pictures human gait is very different from ideal sinusoidal form that is described in most of the papers. So, although we could achieve perfect sinusoidal pathway of the pelvis, it still doesn’t mean that it is natural for every person. In case of human gait, generalization is still necessary for simulating natural and functional gait parameters.

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