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(71) Applicant: NORTHEASTERN UNIVERSITY [US/US];
360 Huntington Avenue, 960 Renaissance Park, Boston,
MA 021 15-5000 (US).(72) Inventors: PIETRUSISNKI, Maciej; 18 Bigelow Street,
Cambridge, MA 02139 (US). MAVROIDIS, Constanti-
nos; 898 Massachusetts Avenue, Arlington, MA 02476
(US).(74) Agents: SCOZZAFAVA, Mary, Rose et al; Wilmer Cut-
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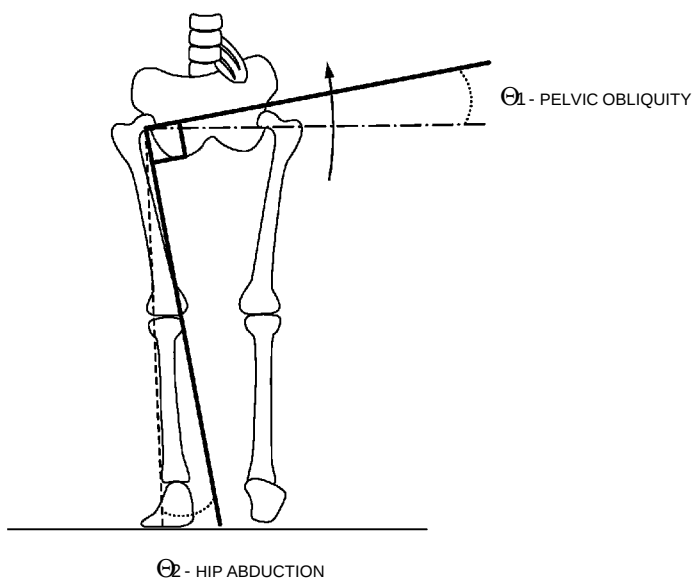
(54) Title: ORTHOPEDIC LOWER BODY EXOSKELETON FOR CONTROL OF PELVIC OBLIQUITY DURING GAIT OVER-
GROUND(57) Abstract: The Over-Ground Robotic Gait Rehabil-
itation Trainer (OG-RGR Trainer) is a wearable and
portable assistive tool for gait rehabilitation and moni-
toring of people with motor control deficits due to a neur-
ological ailment, such as stroke. OG-RGR Trainer cor-
rects secondary gait deviations by applying a corrective
torque around the pelvis, commanded by an impedance
controller. In addition, when featuring actuated knee
flexion/extension, OG-RGR Trainer reinforces a desired
gait pattern by continually applying a corrective torque
around the knee joint, commanded by an impedance
controller. A sensorized yet unactuated brace worn on
the unimpaired leg is used to synchronize the playback
of the desired trajectory based on the user's intent. The
device is mechanically grounded through two ankle foot
orthoses (AFOs) rigidly attached to the main structure,
which helps reduce the weight perceived by the user.

FIG. 14

Orthopedic Lower Body Exoskeleton for Control of Pelvic Obliquity During Gait Over-Ground

CROSS-REFERENCE TO RELATED APPLICATIONS

[0001] This application claims the benefit of U.S. Provisional Patent Application No. 61/567,002, filed on December 5, 2011, which is incorporated herein by reference.

BACKGROUND

Field of Invention

[0002] This invention generally relates to systems for gait retraining. More particularly, the invention relates to a lower extremity exoskeleton designed as a wearable and portable assistive tool for gait neuro-rehabilitation that targets secondary gait deviations (e.g. hip-hiking) by applying force to the pelvis of a patient while walking.

Description of Related Art

[0003] Gait can be defined as a person's particular manner of walking. The upright posture assumed during biped walking is unstable in its nature, and walking is oftentimes described as "a continuous forward fall". Of course, we actually (almost) never fall thanks to the sophisticated coordination of our limbs. Unfortunately, being such a complex process, gait is affected substantially by neurological impairments. According to a 2009 survey, 6.9% of the population in the United States reported an ambulatory disability. The ability to walk and the quality of life are strongly correlated, and therefore restoration of normal gait in the disabled population is of great importance.

[0004] During the past decade, the field of gait rehabilitation has witnessed an increasing interest for the clinical use of robotic systems; particularly in the treatment of neurological ailments such as stroke and traumatic brain injury. Stroke survivors typically receive intensive, hands-on physical and occupational therapy to encourage motor recovery. Manual treadmill locomotor training with partial body weight support (BWS) approach has been proven effective in improving gait of poststroke patients. However, manual treadmill training typically relies on the skill and availability of a physical therapist. Even with the BWS systems, gait training is physically labor intensive. The intensive training required for motor learning is at odds with the availability and cost of a specialized therapist. The scarcity of resources

is exacerbated in cases that require a second, or even a third therapist. For instance, for the retraining of a patient who displays hip hiking due to limited knee flexion, one therapist is needed to guide the knee and another one to control the pelvic obliquity. In such cases, it is an additional challenge for the therapists to maintain coordination.

[0005] Thus there exists a technological gap for a new breed of rehabilitative orthotic devices that accounts for secondary gait deviations while facilitating over ground motion of the patient. In addition, there exists a need for portable gait retraining systems that facilitate over ground locomotion rather than just serving as assistive devices for activities of daily living.

BRIEF SUMMARY

[0006] Methods and systems for providing portable gait retraining systems that account for secondary gait deviations and facilitate over ground locomotion are provided. A wearable and portable assistive tool for gait neuro-rehabilitation that targets secondary gait deviations (e.g., hip-hiking) by reinforcing a desired gait pattern via corrective torque-fields applied around the pelvis is described.

[0007] In one embodiment a gait rehabilitation device is described and, among other components, comprises a pelvic brace, a first leg brace, a second left brace, an impedance controller, and a torque application device for applying a torque onto the pelvis of the user in a frontal plane to counter pelvic obliquity of the user's hip. The first leg brace is rotatably connected to the pelvic brace and includes a first thigh brace and a first shank brace joined by a first rotatable knee joint to provide flexion and extension of the first leg brace. The second leg brace is rotatably connected to the pelvic brace and includes a second thigh brace and a second shank brace joined by a second rotatable knee joint to provide flexion and extension of the second leg brace. An impedance controller computes a desired pelvic obliquity angle based in part on a measured hip abduction angle and commands a torque based in part on the desired pelvic obliquity angle.

[0008] In another embodiment a computer generates a desired pelvic obliquity angle estimate by measuring at least one gait related physiological parameter of a patient, estimating the gait phase based on at least the one gait related physiological parameter, estimating the cadence based on at least the one gait related physiological parameter, selecting a new reference gait trajectory based on at least the estimated cadence; and computing a desired pelvic obliquity angle based on the estimated gait phase and the new reference trajectory.

[0009] These and other features will become readily apparent from the following detailed description where embodiments of the invention are shown and described by way of illustration.

BRIEF DESCRIPTION OF THE SEVERAL VIEWS OF THE DRAWINGS

[0010] For a more complete understanding of various embodiments of the present invention, reference is now made to the following descriptions taken in connection with the accompanying drawings in which:

[0011] Figure 1 shows the phases and subphases of gait;

[0012] Figure 2 is a schematic illustration of the frontal, sagittal and transverse planes through a human body;

[0013] Figure 3 illustrates pelvic drop, anterior tilt, and rotation of the pelvis in normal gait;

[0014] Figure 4 shows the mean trajectories of the angle, velocity, and power of the hip, knee, and ankle joints as a function of gait cycle;

[0015] Figure 5 illustrates hip-hiking of the pelvis;

[0016] Figure 6 illustrates circumduction of the pelvis;

[0017] Figure 7 illustrates a lower extremity exoskeleton for use in secondary gait rehabilitation according to one or more embodiments;

[0018] Figure 8 is a schematic illustration of an application of a moment applied to the pelvis according to one or more embodiments;

[0019] Figure 9 is a schematic view of a pelvic brace illustrating correspondence with a user's hip joints and two options for application of torque according to one or more embodiments;

[0020] Figure 10 is a schematic illustration of a generic impedance control architecture according to one or more embodiments;

[0021] Figure 11 is a schematic illustration of a PD controller acting on pelvic obliquity error according to one or more embodiments;

[0022] Figure 12 is a schematic illustration of the PD gain block of Fig. 11 according to one or more embodiments;

[0023] Figure 13 illustrates computing the pelvic obliquity based on the relative heights of the actuator attachment points;

[0024] Figure 14 illustrates the relationship between the hip abduction angle and the pelvic obliquity angle;

[0025] Figure 15 illustrates mean pelvic obliquity and hip abduction trajectories and the angular deviation from a baseline trajectory during hip-hiking;

[0026] Figure 16 illustrates a synchronization algorithm according to one or more embodiments;

[0027] Figure 17 is a schematic illustration of the system architecture of one embodiment according to one or more embodiments;

[0028] Figure 18 illustrates synchronization of the control system with gait over two consecutive gait cycles;

[0029] Figure 19 a patient in a lower body exoskeleton suspended from a trolley riding on a closed loop track according to one or more embodiments; and

[0030] Figure 20 actuator power units suspended from a trolley with Bowden cables extending to the exoskeleton according to one or more embodiments.

DETAILED DESCRIPTION

[0031] It is easy to underestimate the complexity of locomotion, since we walk automatically and with relative ease. Biped gait is a very intricate process that involves the cooperation of several subsystems. From a mechanical point of view, gait is achieved through the coordination of multiple appendages that are actuated by multiarticular muscles that span more than one joint. From a control system point of view, gait involves the dynamic interactions between the central nervous system (CNS) and the peripheral nervous system (PNS).

Gait Cycle and Gait Deviations

[0032] The human leg has more than seven major degrees of freedom (DoFs), actuated by more than fifteen muscle groups. The human hip is a ball and socket joint with 3-DoFs that allow motion in all three anatomical planes. The knee joint can be simplified as 1-DoF joint that moves along the sagittal plane. Its motions are simply defined as knee flexion/extension. The ankle displays dorsiflexion/plantarflexion in the sagittal plane and inversion/eversion in the coronal plane.

[0033] Walking is a quasi-cyclic motion, which can be divided into gait cycles defined as the period between successive heel strikes of the same foot. Generally, the beginning (and the end) of a gait cycle is defined with the ground contact of the same foot. The gait cycle can further be divided into phases and sub-phases. The stance phase is the period where the particular foot is on the ground and supports the body, which constitutes approximately 60% of the gait cycle. The swing phase is where the

leg is carried forward, and covers the remaining 40%. In symmetrical gait, both limbs are in contact with the ground for about 10% of the cycle, which is referred to as double (limb) support. Figure 1 shows the phases and sub-phases of gait.

[0034] During normal gait, the pelvis rotates in three planes: frontal, sagittal, and transverse. See Fig. 2. Rotation of the pelvis in the frontal plane is termed obliquity, rotation in the sagittal plane is termed pelvic tilt, and rotation in the transverse plane is termed pelvic rotation. During single limb support, these rotations happen about the supporting limb's hip joint. Pelvic drop, anterior tilt and rotation are normal events that occur during normal gait in obliquity, pelvic tilt and pelvic rotation, respectively. See Figure 3.

[0035] The gait cycle is useful in studying gait patterns independently from the variations in timing, since it is defined by events rather than time. Variations in joint parameters throughout a complete gait cycle are typically represented by gait trajectories. Even though the gait cycle representation is normalized for time, the gait trajectories are actually influenced by variations in time dependent parameters, such as walking speed and cadence. Figure 4 shows the mean trajectories of the angle, velocity, and power of the hip, knee, and ankle joints presented as a function of gait cycle (%). Trajectories are shown for six discrete walking velocities ranging from 1 km/hr to 6 km/hr.

[0036] Humans achieve energy-efficiency during gait by using gravity as the main driving force. There is a continuous exchange of kinetic and potential energy, as well as energy exchange between different muscle groups. These synergies can easily be compromised by the changes in timing characteristics following a brain or spinal cord injury. Abnormal synergy patterns arise due to lack of control over individual muscle groups. Unintentional co-contraction of antagonistic muscles may also lead to abnormal torque generation at the joints. As a consequence of these changes in the neuromuscular system, various gait abnormalities take place. Reduced comfortable walking speed (CWS) and/or cadence are common in ambulatory patients. In addition, for fear of falling, longer stance phase of the unimpaired leg can also be observed. This results in asymmetrical gait patterns that are inefficient.

[0037] The deviations from a healthy gait pattern can be classified into two major groups: primary gait deviations and secondary gait deviations. Primary gait deviations are a direct consequence of the underlying impairment and include stiff-legged gait and drop foot. Stiff legged gait is defined as reduced knee flexion during

swing, and is a common primary gait deviation observed in ambulatory patients. The limited toe-clearance poses the risk of tripping over. Stiff-legged gait leads to a gait that is inefficient, unaesthetic, and discomforting. Drop foot is the inability to lift the foot. The ankle flexor muscles are responsible for this motion (or lack thereof). It may cause failure to clear the ground during swing, or result in slapping the ground during heel-strike.

[0038] Secondary gait deviations develop as individuals compensate for their primary gait deviations. These compensatory strategies inevitably cause substantial alterations in their gait patterns. For example, limited flexion of the knee and/or the ankle results in insufficient toe-clearance during swing. Secondary gait deviations include hip hiking and hip circumduction.

[0039] The most common primary gait deviation in patients post-stroke is stiff-legged gait. This gait deviation often results in the patient employing secondary gait deviations that involve motor control of the pelvis. Stiff legged gait is associated with hip hiking or circumduction. Hip hiking is an exaggerated elevation of the pelvis on the contra-lateral side (i.e. hemiparetic side) to allow toe clearance during swing. See Fig. 5. Circumduction is an exaggerated rotation of the pelvis in combination with an exaggerated hip abduction. See Fig. 6. In essence, the swinging of the leg in an arc, as opposed to a motion confined to the sagittal plane. Abnormal control of pelvic obliquity and rotation of the pelvis are the most common secondary gait deviations observed in post-stroke patients. A patient employs these secondary gait deviations in order to assist in foot clearance when either hip flexion or knee flexion is inadequate.

Gait Rehabilitation Systems

[0040] Considering the physical effort involved in such exercises where therapists continually guide the legs and the torso of the patient, robotic neurorehabilitation devices present a great potential as an assistive tool for clinicians by reducing their physical burden. Indeed, some of these systems have already been adopted in clinical practice. Other advantages of robotic systems when compared to manual physical therapy include higher precision and repeatability, and quantitative monitoring of patient's progress via sensors. These factors result in faster and greater level of functional recovery, thus leading to an improvement in patient's level of independence and quality of life.

[0041] Maintaining stability during gait is a major concern for most ambulatory patients. In such cases, they are often prescribed stance-control orthotic braces to

improve stability. However, conventional orthotic braces such as ankle foot orthoses (AFOs) or knee orthoses (KAFOs) typically address the problem of stability by limiting the patient's range of motion (RoM). Such limitation consequentially instigates abnormal gait patterns. For instance, a KAFO designed to increase stability by limiting knee flexion would result in stiff-legged gait. Because of this primary gait deviation, foot clearance would be compromised during the swing phase.

Consequently, the patient would develop compensatory strategies such as hip-hiking and/or circumduction (i.e. secondary gait deviations) to provide ground clearance for the foot. Due to their negative impact on gait patterns, the use of conventional orthotic braces is limited to cases where maintaining stability holds a higher priority than restoring healthy gait patterns. On the other hand, robotic knee orthoses have the potential to overcome the aforementioned limitations by facilitating the knee movement instead of restricting it.

[0042] Robotic gait retraining exoskeletons differ from the conventional orthoses at a very fundamental level: robotic gait re-trainers work towards reinforcing a desired gait pattern and reducing the patient's dependence on assistive technologies; whereas traditional orthoses only mask the symptoms. All rehabilitation robots apply forces on the patient's limbs in one way or the other. One such system is a mechanical exoskeleton worn by an operator. Anthropomorphic exoskeletons attempt to mimic the kinematic structure of the human skeleton. As they work in parallel with the user's limbs, mechanical limits can be implemented directly, and the risk of collisions is eliminated. However, the joints should be accurately aligned with that of the user to prevent shear forces. In joints with a single degree of freedom (DoF), such as the knee joint, it is only a matter of adjusting the exoskeleton limb lengths.

Unfortunately, proper alignment is not as straightforward in joints that have more than 1-DoF. For example, the human hip comprises a ball and socket joint that is located inside the body. Since it is not physically possible to coincide the exoskeleton's joints with that of the patient's other mechanical solutions are required. Some approaches include using a remote center of rotation (CoR) design. Other designs employ a cam and roller mechanism that automatically adjusts its length to compensate for the misaligned abduction/adduction joints while still transferring vertical loads.

[0043] Several robotic devices for gait retraining of stroke patients have been developed in the last decade. The Lokomat (Hocoma AG, Switzerland) is a exoskeletal bilateral gait rehabilitation robot with a BWS system. The patient's legs

are actuated in the sagittal plane via DC motors coupled to ball screws. A spring-based passive foot lifter helps with ankle dorsiflexion during swing. However, the pelvis is only allowed to translate in the frontal plane. Its therapy methods rely on repetition and task-oriented training.

[0044] In contrast to Lokomat, LOPES (LOwer Extremity Powered ExoSkeleton) (University of Twente in the Netherlands) is an example of a new breed of rehabilitation robot that is designed to display low mechanical impedance. Low mechanical impedance is achieved via Bowden cable driven joints that utilize series elastic elements. Series elastic elements add mechanical compliance to the system, which renders higher force-feedback gains possible. In addition to the hip and knee joints that Lokomat controls in the sagittal plane, LOPES features additional actuation of the pelvis in the horizontal plane, as well as hip abduction/adduction. These additional DoF are important, since it is known that fixating the pelvis affects natural walking. All motions of the ankle, and vertical translation of the pelvis are allowed, but are not actuated.

[0045] The aforementioned devices for robotic-assisted gait training neglect gait deviations associated with an abnormal control of the pelvis. The Pelvic Assist Manipulator (PAM) is one robotic device that attempts to address such gait deviations. PAM is described in D. Aoyagi, W. E. Ichinose, S. J. Harkema, D. J. Reinkensmeyer, and J. E. Bobrow, "A robot and control algorithm that can synchronously assist in naturalistic motion during body-weight-supported gait training following neurologic injury," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 15, pp. 387-400, 2007, which is incorporated herein by reference. PAM and other known gait retraining systems that account for secondary deviations are treadmill-based and do not facilitate over ground locomotion. These devices apply corrective force fields to the pelvis via manipulators that are grounded, immovable, and have a limited range of motion. In addition the devices apply corrective force fields to the pelvis based on the angular displacement of the pelvis with respect to the horizontal plane (e.g., the ground).

Over-Ground Robotic Gait Rehabilitation Trainer

The Over-Ground Robotic Gait Rehabilitation Trainer (OG-RGR Trainer) is a wearable and portable assistive tool for gait rehabilitation and monitoring of people with motor control deficits, for example, due to a neurological ailment, such as stroke. OG-RGR Trainer corrects secondary gait deviations by applying a corrective torque

around the pelvis, commanded by an impedance controller. The control algorithm is based, in part, on a measurement of the patient's hip abduction angle as a proxy for the patient's pelvic obliquity angle. The hip abduction angle is not measured with respect to the ground, unlike the pelvic obliquity angle, and can be measured while the patient is walking over the ground. This means that the patient is not restricted to stationary / grounded devices such as treadmills. A sensorized yet unactuated brace worn on the unimpaired leg is used to determine the user's gait over the ground and allows for the synchronization the application of the corrective torque with the relevant portion of the user's gait. The OG-RGR Trainer is mechanically grounded through two ankle foot orthoses (AFOs) rigidly attached to the main structure, which helps reduce the weight perceived by the user. Optionally, when combined with an appropriate active lower extremity exoskeleton, the OG-RGR Trainer reinforces a desired gait pattern by continually applying a corrective torque around the knee joint, also commanded by an impedance controller.

[0046] One embodiment of an OG-RGR Trainer system 70 is illustrated in Fig. 7. The OG-RGR Trainer system includes a pelvic brace 701 and an actuation system 702. The actuation system 702 follows the natural motions of the subject's pelvis, while applying corrective moments to pelvic obliquity, e.g., rotation of the pelvis in the frontal plane, as determined by a control system (not shown). The remaining two rotational DOFs (pelvic rotation (rotation of the pelvis in the sagittal plane) and pelvic tilt (rotation of the pelvis in the transverse plane) and three translational DOFs are non-actuated. The actuation system 702 operates in conjunction with an impedance control system incorporating backdrivability. The control system, described further below, is able to modulate the forces applied onto the body depending on the patient's efforts.

OG-RGR Trainer Exoskeleton

[0047] Due to the presence of soft tissue and the lack of prominent skeletal features in the pelvic region, transfer of torques to alter the orientation of the pelvis and measurement of its orientation in space are challenging. For example, the soft tissue around the pelvic region undergoes significant deformation when a torque is applied to a belt tightened around the waist or pelvis. Also, the applied torques can cause migration, or slipping, of a belt relative to the skin around the pelvic region.

[0048] In one embodiment the OG-RGR Trainer system 70 addresses these challenges by providing a lower extremity exoskeleton 703 that improves torque

transmission to the pelvic region by employing not only adherence to the pelvic region, but also adherence to the thighs, shanks and feet of a patient. Migration of the pelvic brace 701 is substantially eliminated due to the use of the patient's feet, which are positioned transversely to the action of the applied forces of interest, for anchoring the brace to the body. Alteration of the patient's gait is minimal due to the design of the exoskeleton's hip joints, which allow for hip flexion/extension and abduction/adduction, while still transferring forces through the hip joints to the pelvis. This interface maximizes the effectiveness of force transfer to the pelvis, while minimizing time and effort necessary to don and doff the system.

[0049] The lower extremity exoskeleton 703 employs the waist, thighs, shanks and feet of the user to effectively and reliably impart significant forces onto the user's lower body and alter the orientation of the pelvis in the frontal plane (pelvic obliquity). The lower extremity exoskeleton 703 consists of two separate leg braces 704 and 705. Each leg brace includes a thigh portion 706, a shank portion 707, and an ankle foot orthosis (AFO) 708. The thigh portion 706 is attachable to a user's thigh in any suitable manner, such as with a thigh strap. Similarly, the shank portion 707 is attachable to a user's shank in any suitable manner, such as with a shank strap. The thigh portion 706 and shank portion 707 are connected with a rotatable joint 709. The shank portion 707 and AFO 708 are also connected with a rotatable joint 709. Each brace 704 and 705 is attached to the pelvic brace 701 through a 2-DoF hip articulation 710 that creates a remote center of rotation which coincides with the user's hip joint and allows for abduction/adduction and flexion/extension of the hip joint. Each of the joint assemblies (709 and 710) that link portions of the pelvic and exoskeleton together can include sensors, such as rotatable potentiometers 711, for taking measurements of the joint angles.

[0050] The OG-RGR Trainer system employs a single actuation system 702 on one side of the body depending on the particular side of the pelvis that is experiencing a secondary gait deviation. Specifically, the actuation system 702 allows for application of a corrective torque to the pelvis during hip-hike, by controlling the hip abduction-adduction of the ipsi-lateral side (the "healthy" side) when the ipsi-lateral leg is in stance phase, and on the ground. In another embodiment, the actuation system 702 allows for discouraging circumduction, which is another common secondary gait deviation, by generating torque about the hip abduction-adduction of the contra-lateral side (the side affected by stroke) when the leg on the side affected

by stroke is in swing. In another embodiment corrective torques are applied to both hip joints to correct both hip-hiking and circumduction.

[0051] In one embodiment, the control system (described further below) activates the force field only when the leg on the affected side (the hemiparetic leg) is believed to be in swing. This makes it possible to use only one actuator to generate a well-defined moment around the pelvis in the frontal plane, with a vertical reaction force at the support leg, which is equal in magnitude to the applied force generated by the actuator. In one embodiment, the OG-RGR Trainer system uses a synchronization algorithm, discussed further below, which produces an estimate of the subject's location in their own gait cycle, to control the timing of the actuator.

[0052] Figure 8 shows that a single force with an appropriately chosen line of action and proper timing of force application can exert a fully controllable moment onto the pelvis in the frontal plane, provided that the person is in single stance. Specifically, the moment arm **801** consists of a line segment **802** perpendicular to the line of action of the applied force **803** and spanning between it and the hip joint **804** of the supporting leg **805**. See Fig. 8. Figure 9 shows two additional options for the location and mode of applying the required torque directly to the hip joint. In one embodiment, the torque can be applied using an electric motor **902** attached to the pelvic brace **901**. In another embodiment, the torque can be applied from a remote power source using two opposing Bowden cables **903** in a pull-pull arrangement around the rotatable joint. Additional methods of applying the required torque include any means that does not restrict the motion of the patient and including conventional pneumatic and hydraulic actuators. Figure 9 also show the use a spring and cable arrangement **904** that is used to measure the applied torque. As the joint rotates, the cable compresses a spring. The compression of the spring provides a measurement of the torque and is used as part of the feedback loop controlling the system.

Impedance Controller

[0053] The control strategies of rehabilitation robots differ significantly from that of conventional industrial manipulators. One major distinction is that the environment that the robot interacts is a human, with varying system dynamics. Another difference is due to the recent trend in patient-centered exercise regimens, where the robot adapts its behavior based on the patient's effort.

[0054] Mechanical impedance (or simply impedance) is defined as the force response of a system to an imposed motion. Loosely speaking, impedance is the dynamic generalization of stiffness, and is defined as a complex function

$$Z(\omega) = \frac{F(\omega)}{\dot{x}(\omega)}$$

where F is the force vector, $\dot{x}$ is the velocity vector, Z is the impedance matrix and ω is the angular frequency.

[0055] Instead of defining the target position or force (as in position and force controllers, respectively), impedance control defines the dynamic behavior of the robot; which is independent from the environment it interacts with. Impedance control in this context refers to the control of the end-point impedance of a robot or an actuator. A simple schematic of a generic impedance controller is shown in Fig. 10. Impedance control architecture comprises an inner unity feedback force loop and an outer

unity feedback position loop. The main task of the force loop is to increase backdrivability of the actuator. In that sense, force feedback moves any actuator closer to an ideal source of force. The outer position loop sets the relationship between the position of the end-effector, and the force it exerts. In control theory, this can usually be accomplished with a PD controller 1001, where the proportional term represents virtual spring stiffness, and the derivative term acts like a virtual damper. The proportional and derivative gains (PD) produce a force command that is executed by the force loop with gain G 1002. The system's interaction force with the environment (F_{ext} 1003) is measured with a load cell.

[0056] To apply impedance control at the hip abduction level, the control algorithm from the linear-motion / position case discussed above can be adapted to act on angular position error measured in degrees of the hip abduction angle. This system's block diagram is presented in Fig. 11, and the details of the PD controller block are shown in Fig. 12. The strength of the force field is specified with the proportional gain K_c , with units of N-m/deg. For convenience, the derivative gain B_e is not specified independently, but is computed based on the damping ratio, using standard procedures known in the art.

[0057] This type of approach to gain selection allows fast changes to be made to the force - field strength while the general dynamic properties of the system remain unchanged. The PD gains produce a force command, which is executed by the impedance controller's force control loop. Referring to Fig. 11, the PD controller acts

on the hip abduction error and outputs the appropriate force command. Low pass filters 1 and 2 are RC anti-alias filters. Fig. 12 illustrates details of the PD gain block from Fig. 11 when a linear actuator is used. The proportional gain K_c is specified at the hip abduction level, while the derivative gain B_e acts on the linear velocity error at the actuator level. B_e is computed from K_c (the linear motion equivalent) and the specified damping ratio, as would be known in the art. The velocity feedback undergoes secondary filtering (after the velocity error is computed).

Relationship between the Pelvic Obliquity Angle and Hip Abduction Angle

[0058] As noted above, known devices apply corrective force fields to the pelvis via manipulators that are grounded, immovable, and have a limited range of motion. In addition, the devices apply corrective force fields to the pelvis based on the angular displacement of the pelvis with respect to the horizontal plane (e.g., the ground).

[0059] For example, a grounded device calculates the pelvic obliquity angle of the pelvic brace using the relative position of the two attachment points on the pelvic brace (in the vertical direction) and the distance between these two points. See Figure 13. As one side of the pelvis moves upward with respect to the other, the segment of length D spanning the two attachment points rotates. If this segment " D " serves as the hypotenuse of a right triangle and the difference in height between the two attachment points is " y ", then the resulting angle of rotation " Θ " is the pelvic obliquity angle.

[0060] In the case of a grounded system, the relative vertical position of the attachment points can be computed using the ground as a reference point. However, the pelvic obliquity angle cannot be computed in this manner when the patient is mobile. For example, when the user is walking over the ground and not on a fixed treadmill type device. It has been found that there is a correlation between the pelvic obliquity angle and the hip abduction angle of the leg in stance during single leg . When a patient exhibits hip-hike, their pelvis rotates about the non-affected (contra-lateral) side's hip joint, with the contra-lateral leg in stance, and hence supporting the whole body weight.

[0061] Figure 14 shows the pelvic obliquity angle (as measured with respect to the ground) and the contra-lateral hip abduction angle. Figure 15A shows the mean pelvic obliquity and the mean right hip abduction trajectories through a complete gait cycle (left heel strike to next left heel strike) for both the normal and the hip-hiking cases. Figure 15B shows the pelvic obliquity trajectory difference between the mean

normal and hip-hiking cases. Figure 15B also shows the mean right hip abduction difference between the normal and hip-hiking cases. During the second portion of the gait cycle (when hip-hiking occurs in the presented example) the measured deviation from the nominal trajectory, or position error, for the hip abduction angle is very similar to the measured deviation from the nominal trajectory for the pelvic obliquity angle. The hip abduction angle can therefore be used as a proxy for the pelvic obliquity angle.

[0062] The hip abduction angle can be measured with rotary potentiometer placed at the exoskeleton's hip abduction axis, or with other conventional methods, such as a linear potentiometer or rotary or linear encoder. This means that, when used as a proxy for the pelvic obliquity angle, the hip abduction angle can be used as an input to the impedance controller in systems that are not rigidly connected to the ground. Because the hip abduction angle can be measured without rigid connection to ground, and because torques can be applied onto the pelvis at hip abduction to control pelvic obliquity, it is possible to have mobile, or over the ground, robotic-assisted gait training that addresses secondary gait deviations such as hip-hiking.

Reference Trajectory

[0063] A pelvic obliquity reference trajectory is a time series, containing the relationship between space and time. Therefore, in addition to being properly positioned in space, the individual data points of the reference trajectory also have to be presented to the impedance controller at the right time. Therefore, a synchronization algorithm is implemented in the OG-RGR Trainer. One suitable synchronization algorithm is available at D. Aoyagi, W. E. Ichinose, S. J. Harkema, D. J. Reinkensmeyer, and J. E. Bobrow, "A robot and control algorithm that can synchronously assist in naturalistic motion during body-weight-supported gait training following neurologic injury," IEEE Transactions on Neural Systems and Rehabilitation Engineering, vol. 15, pp. 387-400, 2007 ("the Aoyagi synchronization algorithm"), which is incorporated in its entirety by reference.

[0064] The duration of a single gait cycle spans between two consecutive left heel strikes. The right heel strike occurs at the 50% mark in the gait cycle (assuming symmetrical gait). The Aoyagi synchronization algorithm estimates the actual temporal position of the subject within his gait cycle based on the angular positions and velocities of the subject's hip and knee joints (8 degrees of freedom). A reference for the synchronization algorithm is constructed by recording an 8-dimensional time

series over several gait cycles and finding the normalized mean of each DOF. The 8 DOFs are normalized to ensure that they are assigned equal weight. The reference is generated by the norms of the individual vectors, and is represented by the loop of discrete points in Fig. 16. The number of discrete points in the reference is a function of walking cadence and sampling rate used.

[0065] During operation, a minimization operation of the norm of the difference between the measured 8-dimensional vector and every vector in the reference is performed, and this identifies the location of the nearest neighbor. This result is normalized to give an index value ranging between 0 and 1. This represents the location of the subject in the temporal sense in the gait cycle.

[0066] The lower extremity exoskeleton 703 features knee and hip angle measurements (4 DOFs). Taking derivatives of these signals produces four angular velocities, for a total of 8 DOFs for use in the Aoyagi synchronization algorithm. In addition, a low profile assembly with a micro switch, which is placed in the subject's shoe, is used to detect heel strikes. In one embodiment, the micro switch is mounted on an aluminum sheet sized to fit in the shoe and covered with a plastic sheet for user comfort. Knowledge of such a discrete gait event is useful for both generating synchronization reference trajectories, and for synchronization algorithm performance validation purposes.

[0067] Signals from four rotary potentiometers at the hip and knee joints are analog low pass RC - filtered and sent to a data acquisition card. A heel strike signal, which is also collected, is used to parse the data and find 8 means of the 8 DOFs (hip and knee angular positions and velocities) across the multiple gait cycles.

[0068] The overall control system architecture is illustrated in Fig. 17. This control system includes a first controller built up around the pelvic obliquity impedance controller, discussed above, and includes the Aoyagi gait estimation algorithm discussed above. This control system allows for modulation and fine-tuning of the force field applied on to the user in two ways.

[0069] First, the controller can switch between two (or more if necessary) different position references while in operation, within two consecutive gait cycles. The user's hip and knee joint angular positions and velocities are used by the Aoyagi gait estimation algorithm to produce an estimate of the user's point in the gait cycle at any time. This estimation of the point in gait is used in two lookup tables to generate two position references.

[0070] Switch 1 shown in Fig. 17 executes a transition between the two reference trajectories. This switch follows a sigmoid curve, which is a section of a 3 Hz sinusoid, spanning between 0 and 1. Switch 1 is set to go on or off beginning at 20% of the gait cycle, when the contralateral leg is in stance.

[0071] The second way to control the force field applied onto the subject is through precise activation and de-activation of the impedance gains. Switch 2 in Fig. 17 follows a sigmoid curve as well, enabling a smooth transition from the backdrivable mode (zero force control) to impedance control mode, when the PD gains set the desired stiffness and damping (the force field). Fig. 18 illustrates an example of the operation of Switch 2 over to consecutive gait cycles. The synchronization algorithm output predicts left heel strikes and gives a good estimate of gait cycle location mid-stride. The gait estimation (Synchr Output) is the progression through the gait cycle from 0 to 1 (100%). The force field activation sigmoid switch (3Hz) was set to go on at 44% and off at 76%. Heel strike is marked by the rising edge of the 'Heel Strike Switch' signal.

Additional Embodiments

[0072] In an additional embodiment the OG-RGR Trainer system described above can be combined with a body weight support system. Figure 19 shows a patient in an OG-RGR Trainer suspended from a trolley on a closed loop track. In this embodiment, the OG-RGR Trainer's active joints are actuated with rotary series elastic elements via remote transmission (i.e. push-pull Bowden cables). The actuators generating the torque are located in a unit suspended from the BWS trolley, minimizing the mass worn by the patient. See Figure 20. This allows for the use of an active joint at each hip as well as active control of the knee and ankle without weighing the patient down with multiple motors on the exoskeleton.

CLAIMS

1. A lower extremity exoskeleton for gait rehabilitation comprising:
 - a pelvic brace;
 - a first leg brace rotatably connected to the pelvic brace, further comprising:
 - a first thigh brace and a first shank brace joined by a first rotatable knee joint to provide flexion and extension of the first leg brace, wherein the first thigh brace and first shank brace are adapted to be coupled to the unaffected leg of a user;
 - a second leg brace rotatably connected to the pelvic brace, further comprising:
 - a second thigh brace and a second shank brace joined by a second rotatable knee joint to provide flexion and extension of the second leg brace, wherein the second upper and second lower leg shanks are adapted to be coupled to the affected leg of a user;
 - a first sensor for measuring the hip abduction angle of the user;
 - a torque application device for applying a torque onto the pelvis of the user in a frontal plane to counter pelvic obliquity of the user's hip; and
 - an impedance controller for computing a desired pelvic obliquity angle based in part on the measured hip abduction angle and commanding the torque based in part on the desired pelvic obliquity angle.

2. The system of claim 1 wherein the impedance controller receives at least one gait related physiological parameter of the user; and estimates the cadence of the user based on at least the one gait related physiological parameter.
3. The system of claim 2 wherein the impedance controller selects a reference gait trajectory based on the estimated cadence of the user.
4. The system of claim 1, 2, or 3 further comprising a gait rehabilitation system for addressing primary gait deviations.
5. The system of claim 1, 2, 3, or 4 wherein the impedance controller computes a desired hip abduction angle based on an estimated gait phase of the user and an estimated cadence of the user.
6. The system of claim 1, 2, 3, or 4 wherein the torque application device comprises a linear actuator.
7. The system of claim 1, 2, 3, or 4 wherein the torque application device comprises an electric motor.
8. The system of claim 1, 2, 3, or 4 wherein the torque application device comprises a remote power source and two opposing Bowden cables in a pull-pull arrangement.
9. The system of claim 1, 2, 3, or 4 further comprising a second sensor for measuring knee flexion and extension angles.
10. The system of claim 4 or 5 further comprising means for creating a torque around the axis of rotation of the second rotatable knee joint.
11. The system of any of the preceding claims wherein the impedance controller commands the torque in synchronization with the user's gait.
12. The system of any of the preceding claims wherein the impedance controller commands the torque when the user's leg is in a swing phase.
13. The system of any of the preceding claims wherein the first and second upper and lower leg shanks are slotted for adjusting the distance from hip to the knee joint and from the knee joint to the ankle joint, respectively.

14. The system of any of the preceding claims wherein the leg braces are each connected to the pelvic brace using a spherical joint that allows for flexion/extension, abduction/adduction, and rotation.
15. The system of any of the preceding claims wherein the first leg brace further comprises a sensor for measuring knee flexion and extension angles.
16. The system of any of the preceding claims wherein the sensor includes a rotary potentiometer and a rotary encoder.
17. The system of any of the preceding claims wherein each leg brace further comprises a telescopic structure that allows the exoskeleton structure to shorten during abduction.
18. The system of any of the preceding claims further comprising a first ankle foot orthosis coupled to the first leg brace and a second ankle foot orthosis coupled to the second leg brace.
19. The system of claim 18 wherein the first ankle foot orthosis includes a pressure transducer.
20. The system of claim 18 wherein the first leg brace further comprises a sensor for measuring ankle flexion and extension angles.
21. The system of claim 19 or 20 further comprising a controller for computing a desired knee angle at the first rotatable knee joint based in part on the sensor measurements and commanding a torque at the second rotatable knee joint based in part on the desired knee angle.
22. The system of claim 21 wherein the impedance controller further computes a desired knee joint angle based on an estimated gait phase of the user and an estimated cadence of the user.
23. A gait rehabilitation method comprising:

measuring the hip abduction angle of the user;

computing a desired pelvic obliquity angle and a desired torque based in part on the measured hip abduction angle;

applying the desired torque onto the pelvis of the user in a frontal plane to counter a secondary gait deviation of the user.

24. The method of claim 23 further comprising receiving at least one gait related physiological parameter of the user; and estimating the cadence of the user based on at least the one gait related physiological parameter.
25. The method of claim 24 further comprising selecting a reference gait trajectory based on the estimated cadence of the user.
26. The method of claim 22, 23, or 24 further comprising applying a torque onto a knee of the user to counter the user's primary gait deviations.
27. The method of claim 22, 23, 24, or 25 further comprising computing a desired hip abduction angle based on an estimated gait phase of the user and an estimated cadence of the user.
28. The method of claim 22, 23, 24, or 25 wherein the means for creating a torque comprises a linear actuator.
29. The method of claim 22, 23, 24, or 25 wherein the desired torque is applied using an electric motor.
30. The method of claim 22, 23, 24, or 25 wherein the desired torque is applied using a remote power source and two opposing Bowden cables in a push-pull arrangement.
31. The method of claim 22, 23, 24, or 25 further comprising receiving measurements of knee flexion and extension angles.
32. The method of claim 26 or 27 further comprises creating a torque around the axis of rotation of the second rotatable knee joint.
33. The method of claims 22 through 32 wherein the torque is applied in synchronization with the user's gait.
34. The method of claims 22 through 33 wherein the torque is applied when the user's leg is in a swing phase.

35. The method of claim 31 further comprising computing a desired knee angle at a first rotatable knee joint based in part on the knee measurements and commanding a torque at a second rotatable knee joint based in part on the desired knee angle.
36. The method of claim 35 further comprising computing a desired knee joint angle based on an estimated gait phase of the user and an estimated cadence of the user.

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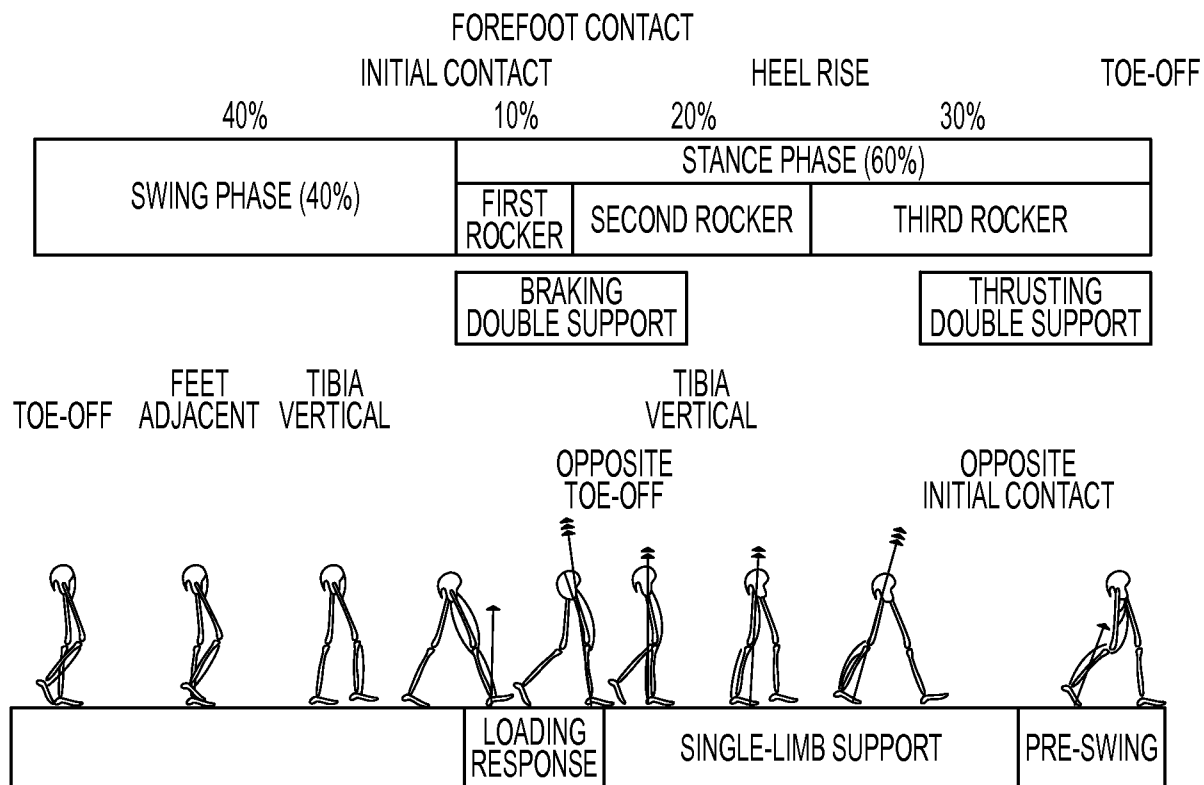


FIG. 1

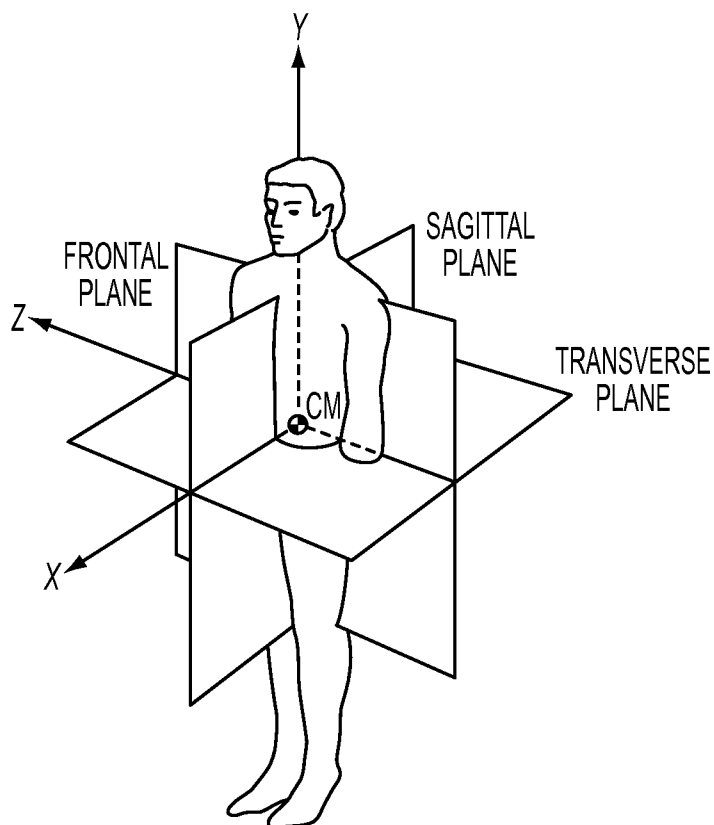


FIG. 2

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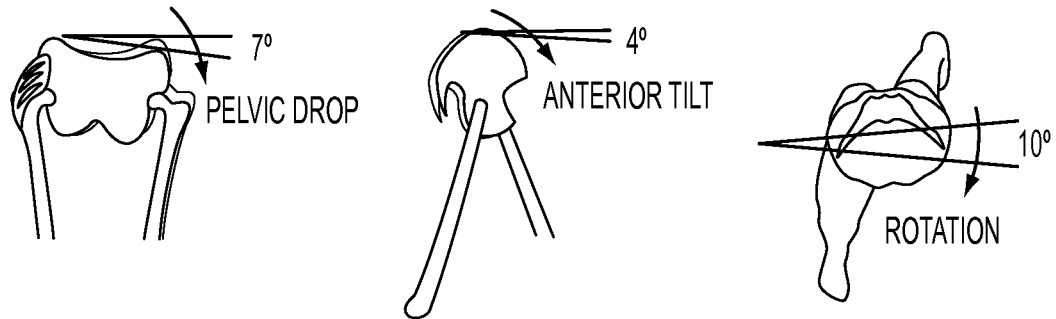


FIG. 3

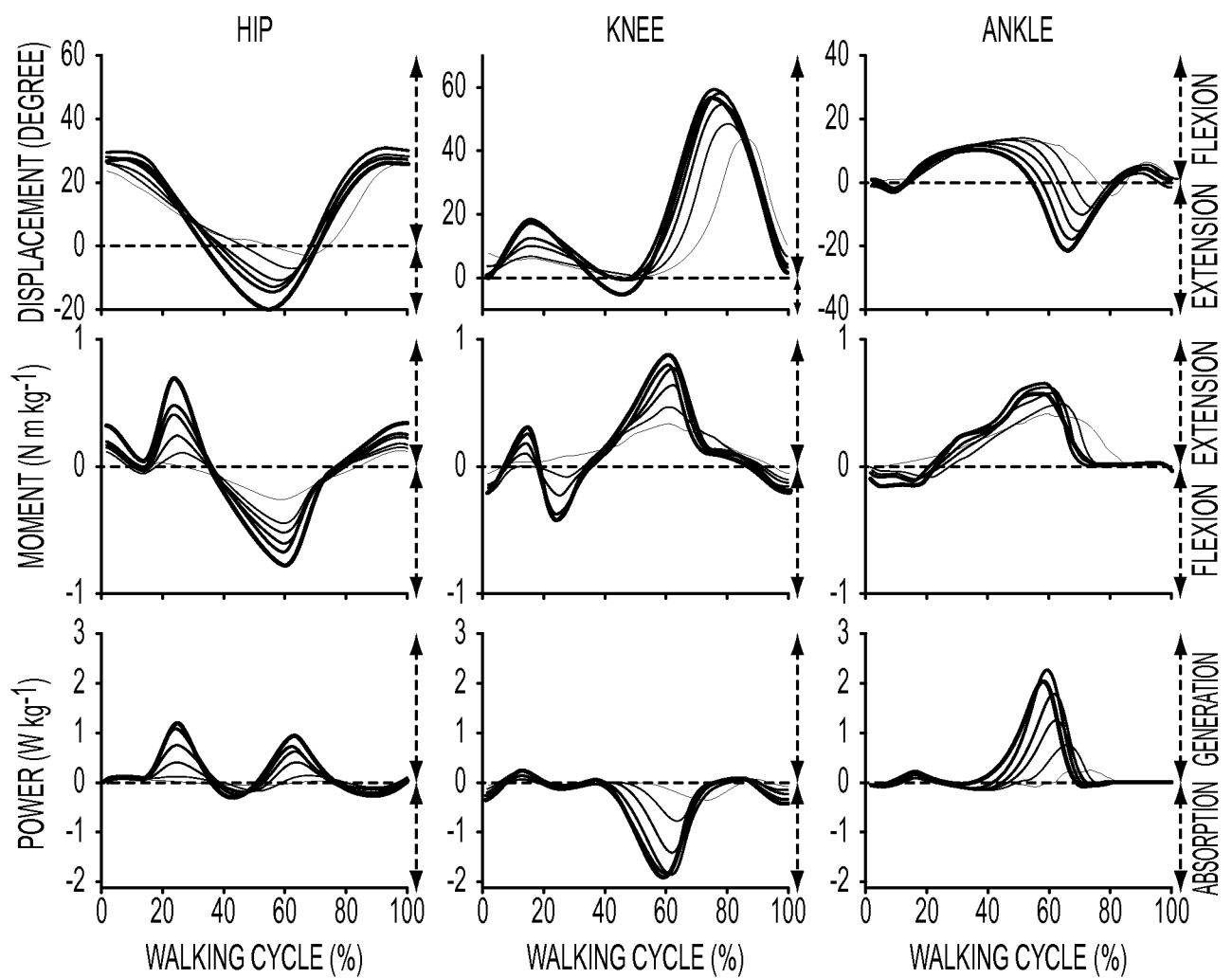


FIG. 4

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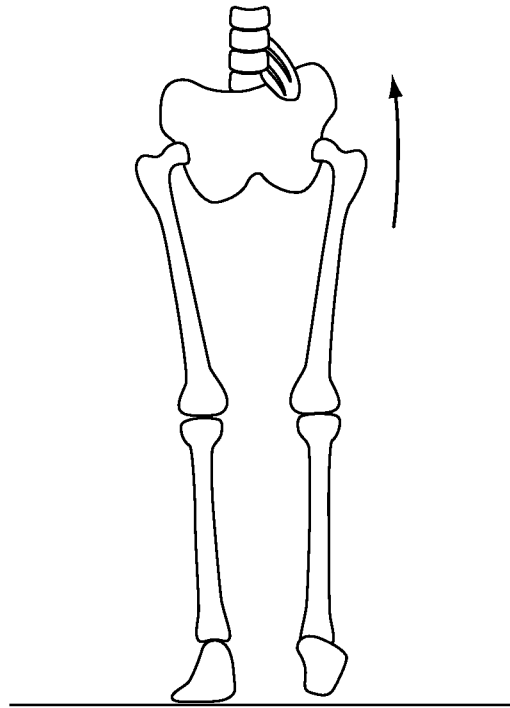


FIG. 5

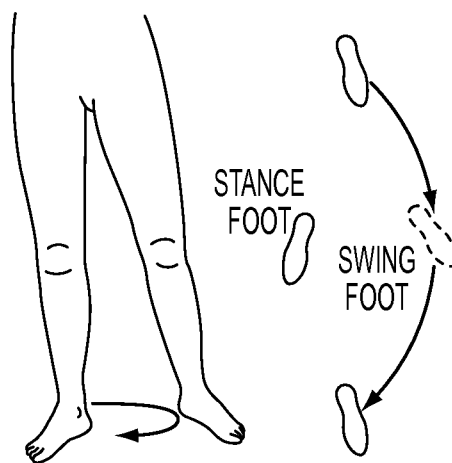


FIG. 6

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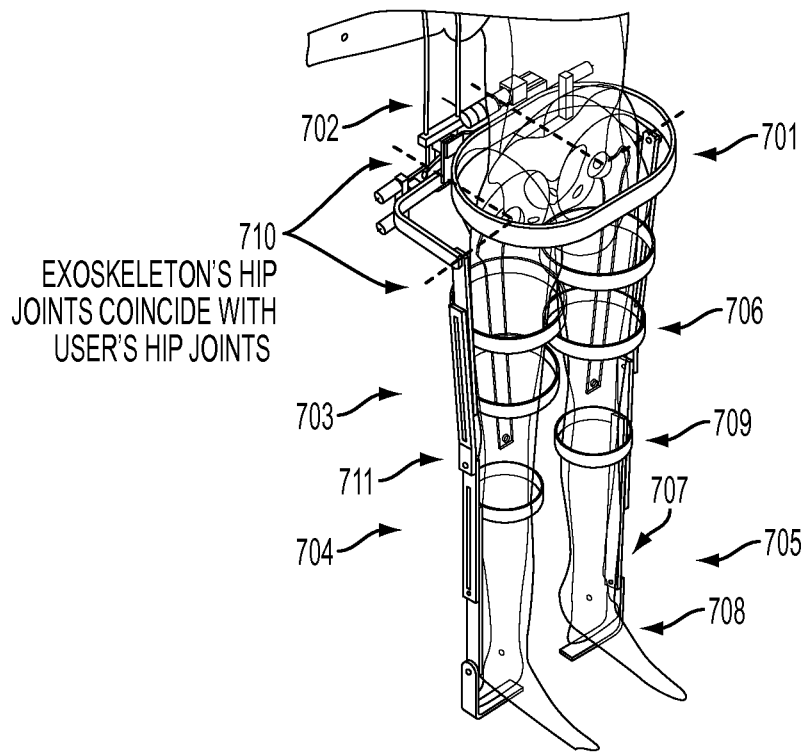


FIG. 7

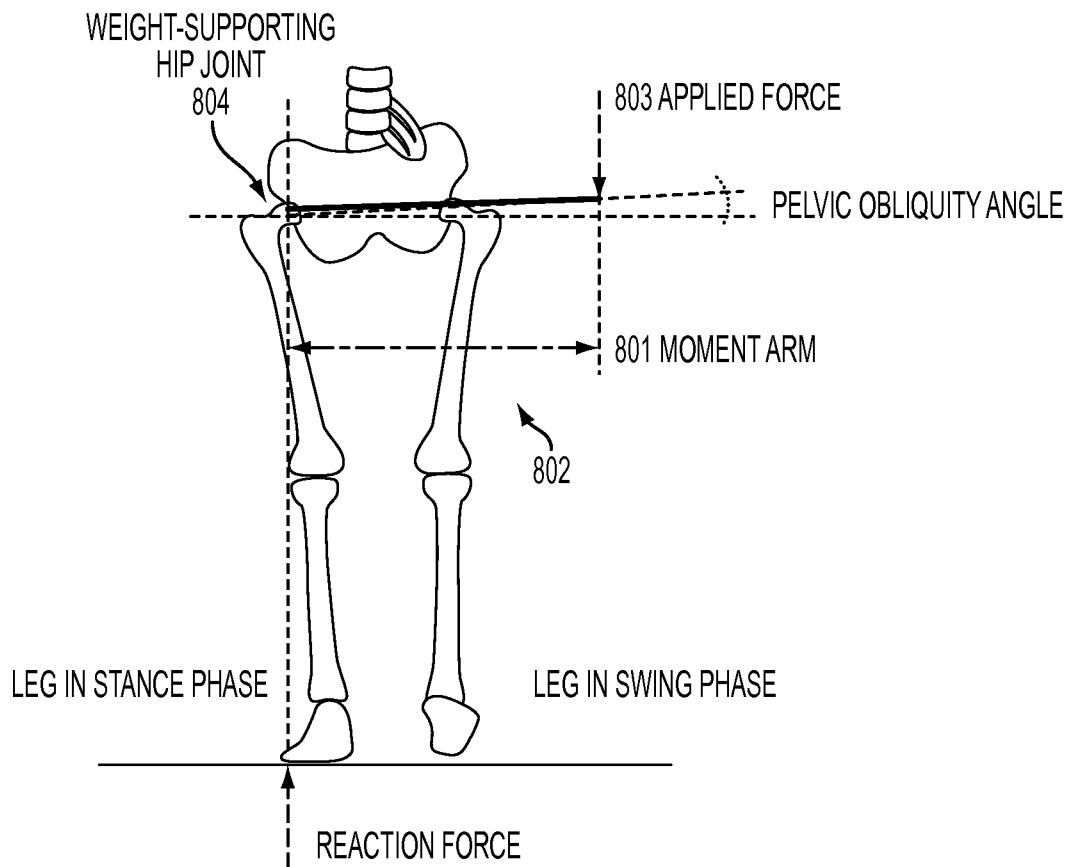


FIG. 8

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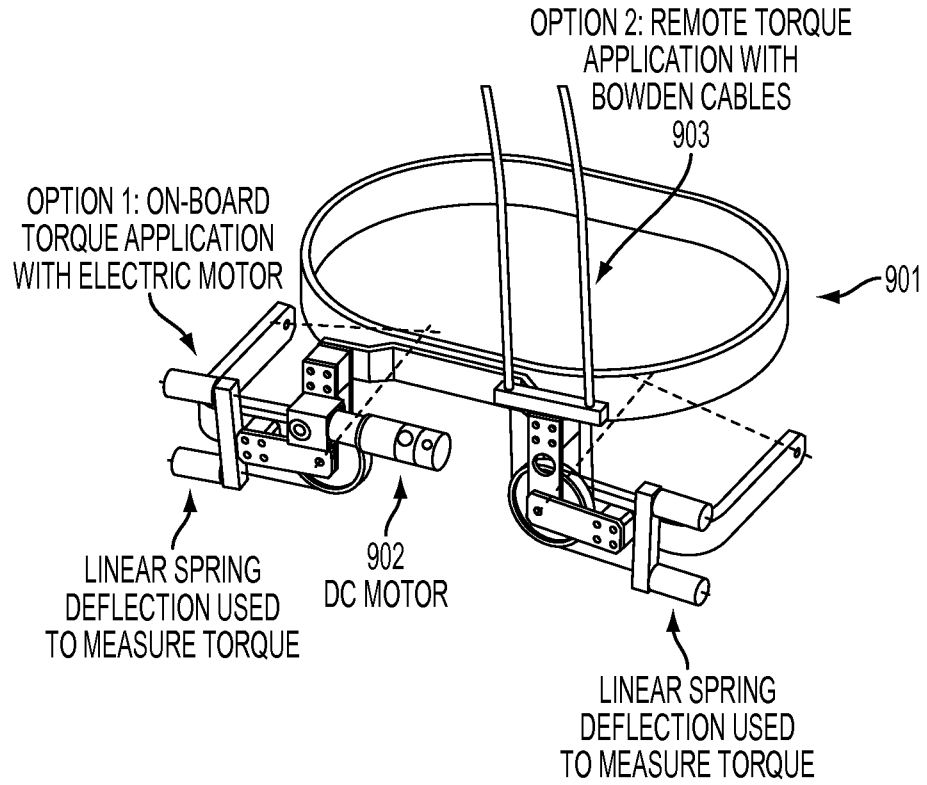


FIG. 9

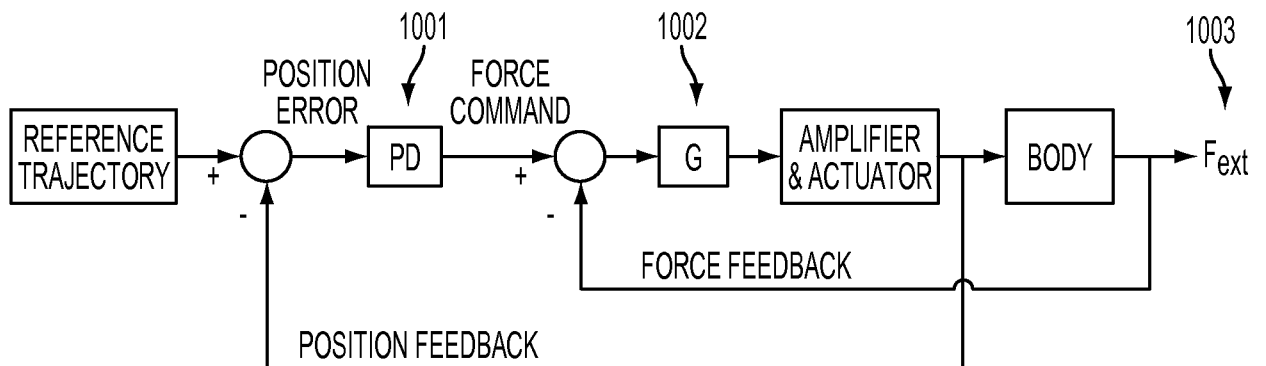


FIG. 10

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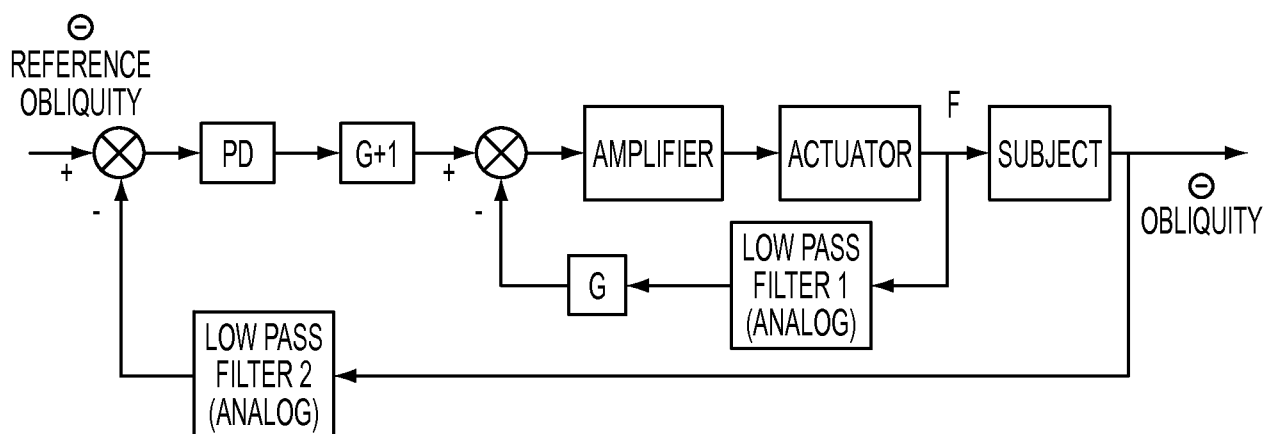


FIG. 11

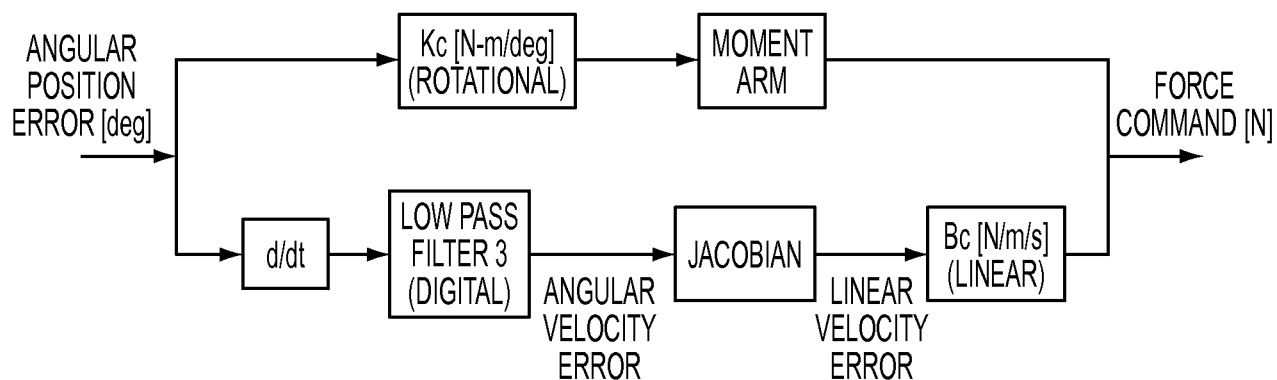


FIG. 12

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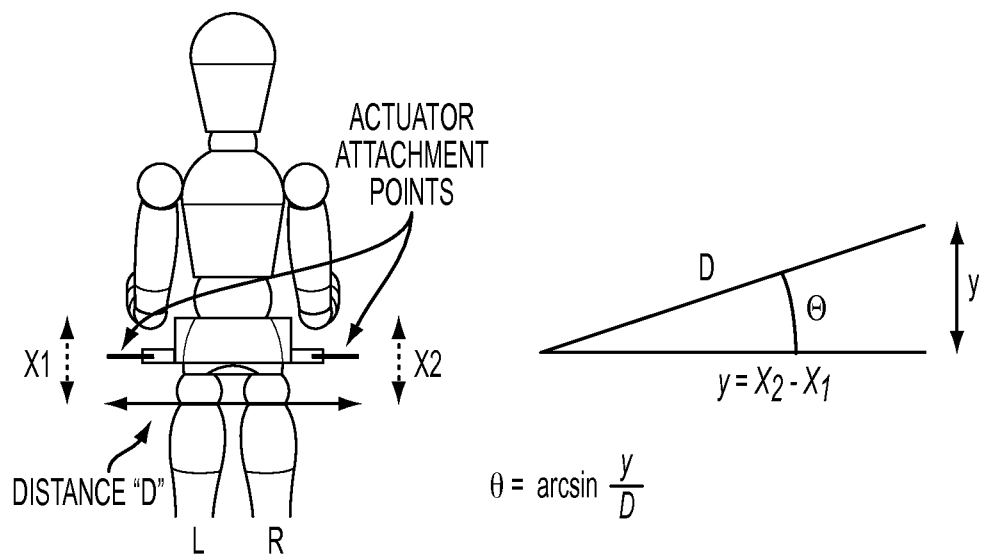


FIG. 13

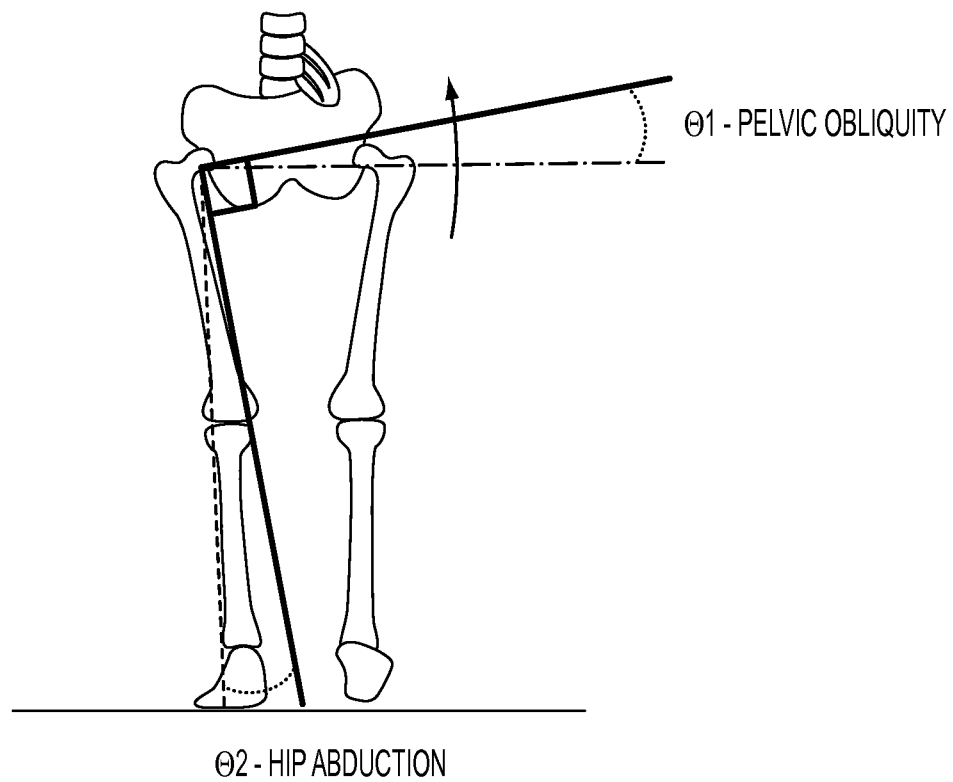


FIG. 14

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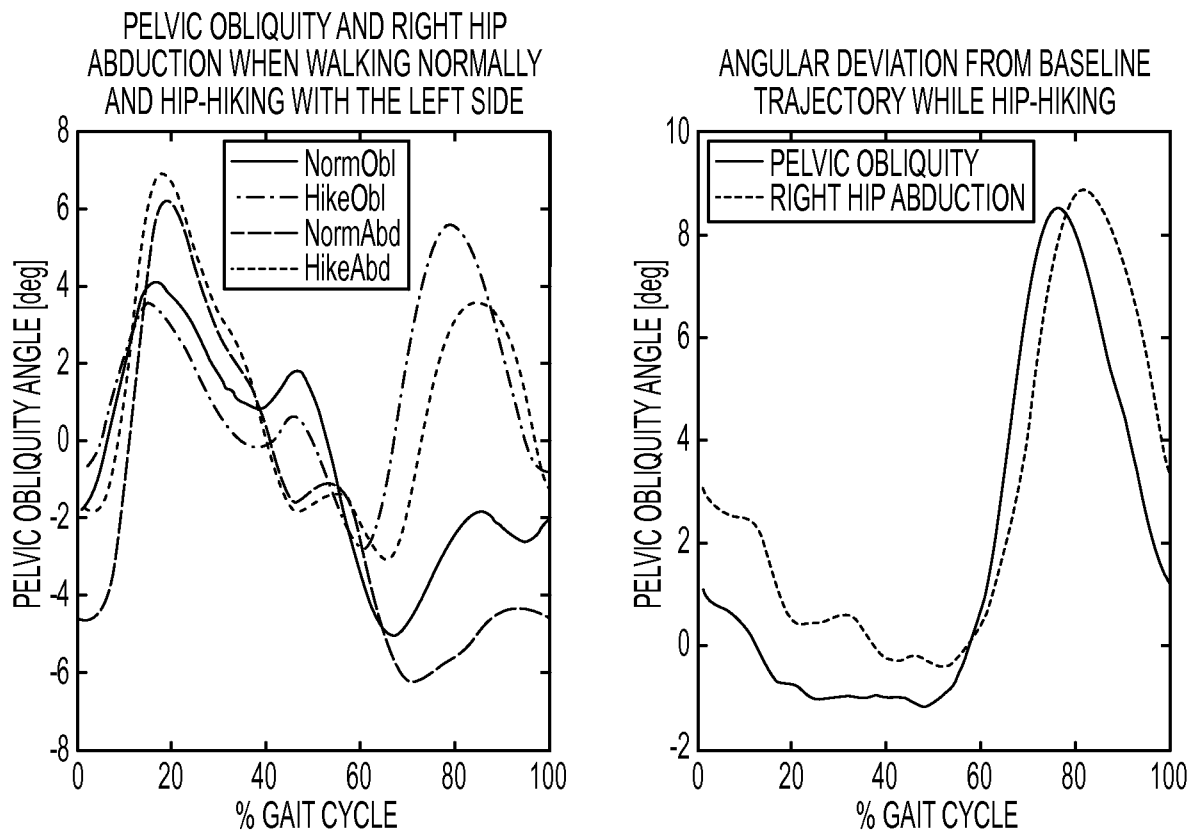


FIG. 15

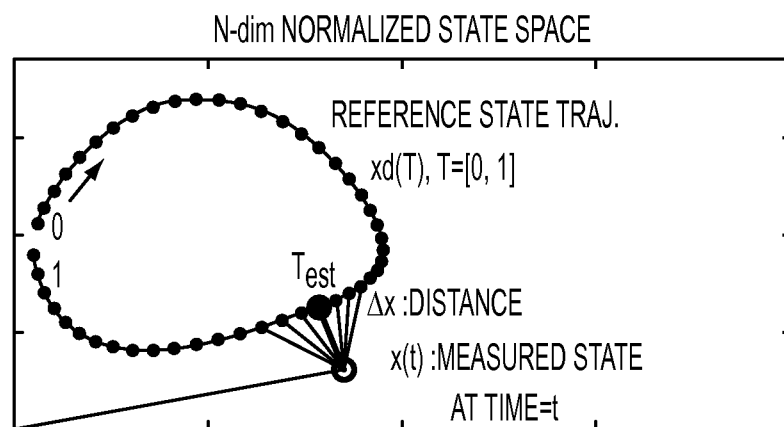


FIG. 16

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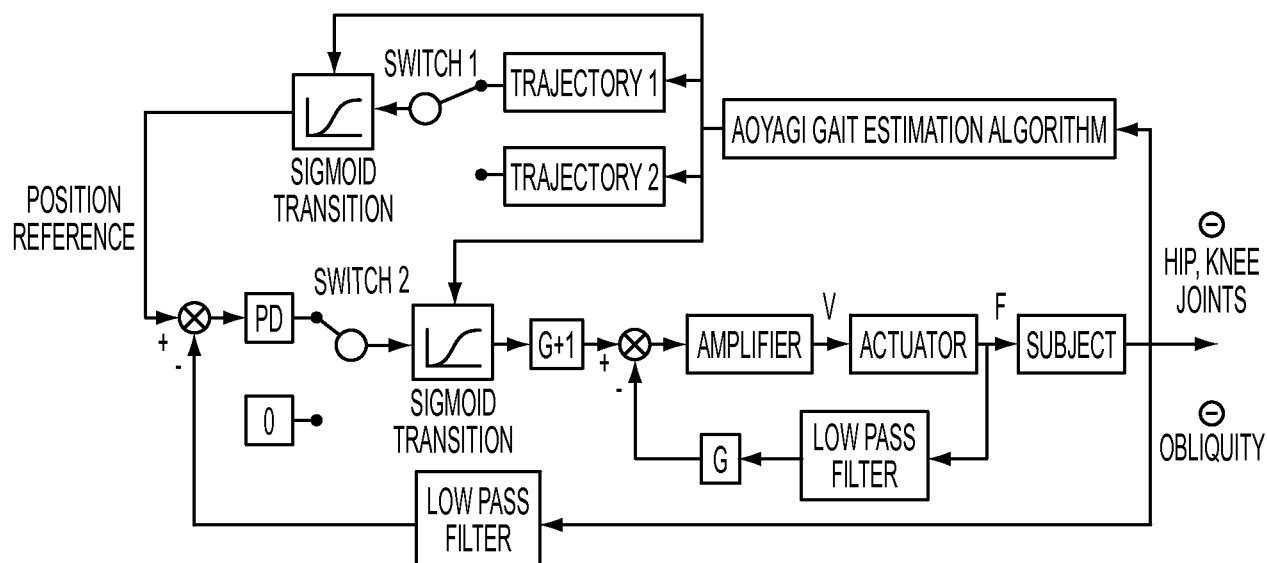


FIG. 17

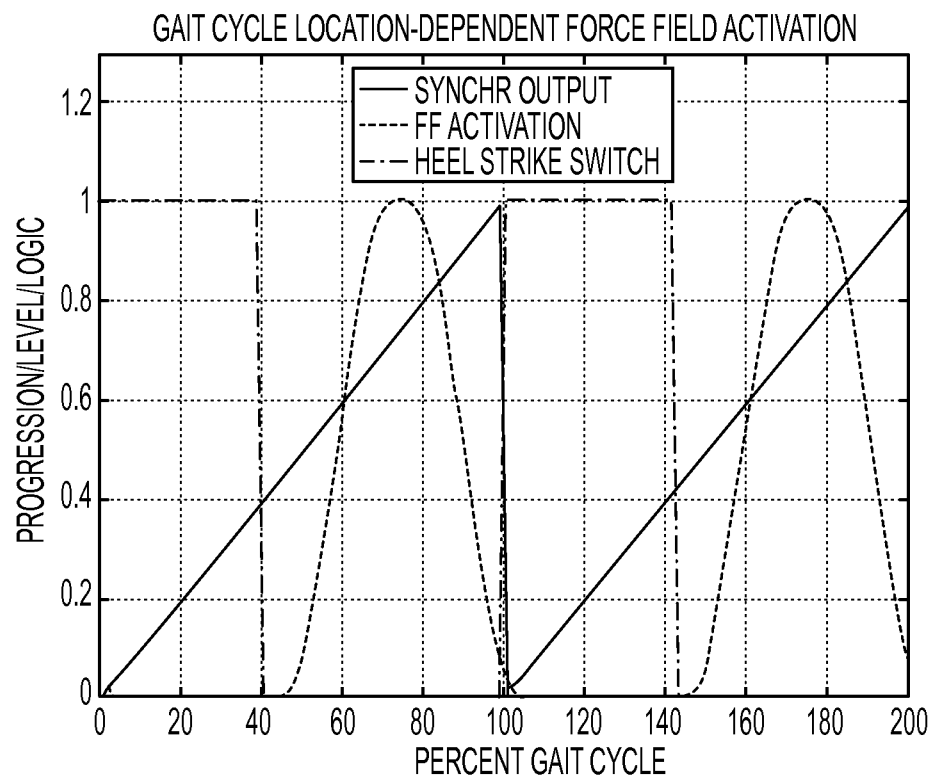


FIG. 18

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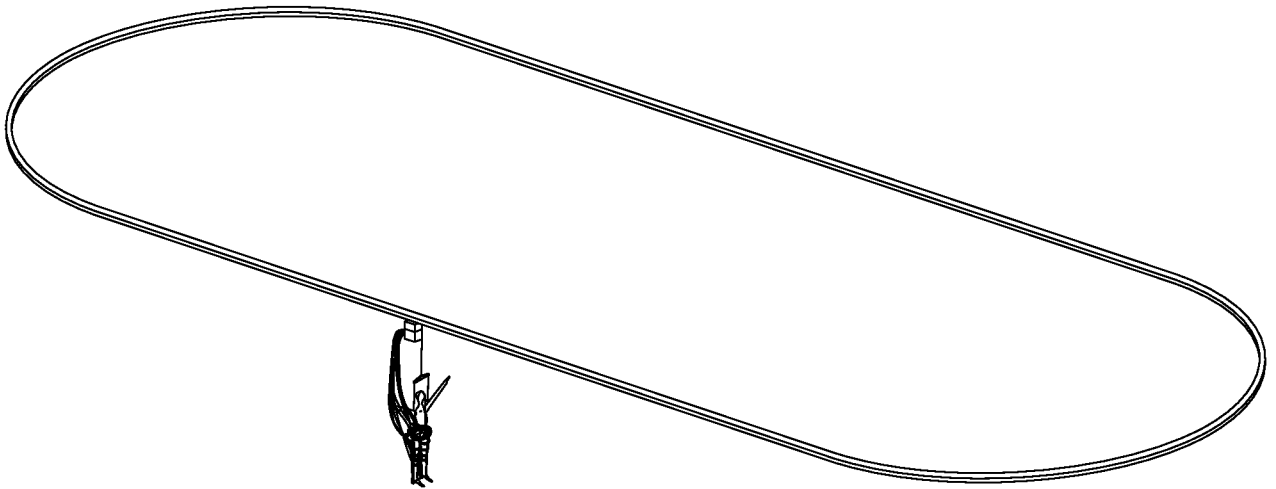


FIG. 19

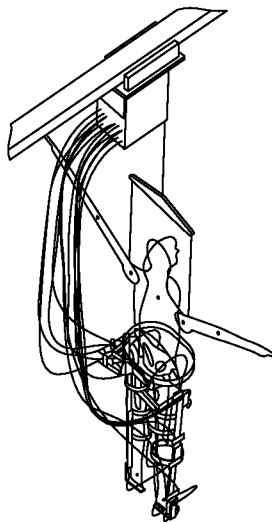


FIG. 20

INTERNATIONAL SEARCH REPORT

International application No.

PCT/US 12/68012

A. CLASSIFICATION OF SUBJECT MATTER

IPC(8) - A61 H 1/00 (2013.01)

USPC - 601/34, 35

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

IPC(8) ? A61H 1/00 (2013.01)

USPC ? 601/34, 35

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

602/23; 482/54, 66, 69

(Search term limited; see below)

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

PubWest (PGPB, USPT, EPAB, JPAB); Google Scholar

Search Terms: Gait, walking, training, rehabilitation, rehab, assist, hip, pelv*, obliquity, frontal rotation, coronal rotation, abduction, adduction, measure, sensor, monitor, transducer, potentiometer, gauge, record

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
Y	(Pietrusinski et al.) Design of Human - Machine Interface and Altering of Pelvic Obliquity with RGR Trainer. IEEE Int Conf Rehabil Robot 201 1 June: 1-6.	1-4, 23-26
Y	US 2009/0024065 A1 (EINARSSON) 22 January 2009 (22.01.2009) Entire document, especially Abstract, para[0018], para[0043]- para[0045], para[0075].	1-4, 23-26
Y	(Aoyagi et al.) A Robot and Control Algorithm Than Can Synchronously Assist in Naturalistic Motion During Body-Weight-Supported Gait Training Following Neurologic Injury. IEEE Transactions On Neural Systems and Rehabilitation Engineering, Vol 15, No. 3 September 2007, p389-400.	4, 26



Further documents are listed in the continuation of Box C.



* Special categories of cited documents:

"A" document defining the general state of the art which is not considered to be of particular relevance

"E" earlier application or patent but published on or after the international filing date

"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art

"&" document member of the same patent family

Date of the actual completion of the international search

24 January 2013 (24.01.2013)

Date of mailing of the international search report

19 FEB 2013

Name and mailing address of the ISA/US

Mail Stop PCT, Attn: ISA/US, Commissioner for Patents
P.O. Box 1450, Alexandria, Virginia 22313-1450

Facsimile No. 571-273-3201

Authorized officer:

Lee W. Young

PCT Helpdesk: 571-272-4300

PCT OSP: 571-272-7774

INTERNATIONAL SEARCH REPORT

International application No.

PCT/US 12/68012

Box No. II Observations where certain claims were found unsearchable (Continuation of item 2 of first sheet)

This international search report has not been established in respect of certain claims under Article 17(2)(a) for the following reasons:

1. ☐ Claims Nos.:
because they relate to subject matter not required to be searched by this Authority, namely:
2. ☐ Claims Nos.:
because they relate to parts of the international application that do not comply with the prescribed requirements to such an extent that no meaningful international search can be carried out, specifically:
3. ☒ Claims Nos.: 5-22, 27-36
because they are dependent claims and are not drafted in accordance with the second and third sentences of Rule 6.4(a).

Box No. III Observations where unity of invention is lacking (Continuation of item 3 of first sheet)

This International Searching Authority found multiple inventions in this international application, as follows:

1. H As all required additional search fees were timely paid by the applicant, this international search report covers all searchable claims.
2. ☐ As all searchable claims could be searched without effort justifying additional fees, this Authority did not invite payment of additional fees.
3. ☐ As only some of the required additional search fees were timely paid by the applicant, this international search report covers only those claims for which fees were paid, specifically claims Nos.:
4. ☐ No required additional search fees were timely paid by the applicant. Consequently, this international search report is restricted to the invention first mentioned in the claims; it is covered by claims Nos.:

Remark on Protest

- ☐ The additional search fees were accompanied by the applicant's protest and, where applicable, the payment of a protest fee.
- ☐ The additional search fees were accompanied by the applicant's protest but the applicable protest fee was not paid within the time limit specified in the invitation.
- ☐ No protest accompanied the payment of additional search fees.