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A Split-Crank, Servomotor-Controlled Bicycle Ergometer Design for Studies in Human Biomechanics

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Abstract

This paper presents a novel computer-controlled bicycle ergometer, the TiltCycle, for use in human biomechanics studies of pedaling. The TiltCycle has a tilting (reclining) seat and backboard, a split crank to isolate the left and right loads to the feet of the cyclist, and two belt-driven, computer-controller motors to provide both assistance and resistance loads. Sensors measure the kinematics and force production of the pedaling work performed, as well as goniometer and electromyography signals from the lower limbs. The technical description includes the mechanical design, low-level software and control algorithms designed for studies in human lower-limb biomechanics and bilateral coordination, and concludes with validation testing and system identification results.

1. Introduction

Scientific studies in human motor control, strength and coordination require both the delivery of stimuli to subjects during exercise and the measurement of kinematic and dynamic variables during the resulting movement. For studies in lower-limb function, it is often desirable to perform studies in settings that closely resemble actual human experience, e.g., studying walking in a Gait Lab with video-based motion analysis and floor-mounted force plates as the prime measurement tools. However, for some studies it is important to be able to constrain certain degrees of freedom of movement and provide repeatable, kinematics-triggered force stimuli. To study human lower-limb coordination and motor control in neurologically intact and impaired subjects, we have developed a split-crank bicycle ergometer with the capability to deliver computer-controlled motion and force stimuli separately to each of the two pedaling legs, and simultaneously measure up to 64 channels of kinematic, dynamic, goniometer and electromyography signals. This system, called the TiltCycle™, is the result of an evolutionary process of development at this VA Rehabilitation R&D Center since the early 1980s. It is one of a number of experimental devices developed at the center to exploit the ability of fast real-time computation to deliver immersive mechatronic environments for the testing of human performance and neuromotor control [1].

1.1 Ergometer Designs

Ergometers are used in sports training and rehabilitation to measure and increase strength and range of motion. Bicycle trainers with floor-level rollers, used in conjunction with a cyclist's own bicycle, as well as stationary bicycles with variable and programmable resistance control, are commercially available for aerobic exercise and rehabilitation (e.g., Monark AB, Varberg, Sweden). For research studies in human biomechanics, several laboratories have developed bicycle ergometers combined with data acquisition systems [e.g., 2, 3, 4]. There are also non-bicycle, lower-extremity, and whole-body exercise systems, such as steppers and rowers, also used in sports and exercise research (e.g., Concept2, Morrisville, VT, USA). The use of motors in ergometers to provide both assistive and resistive loading is growing (e.g., BioDex Exercise Systems, New York, USA). Several upper-extremity systems currently used in biomechanics and sports research include the stroke rehabilitation robots MIME [5], Driver's SEAT [6], MIT-MANUS [7], and the GENTLE system [8]. The Lokomat (Hocoma AG, Zürich, Switzerland) multi-degree-of-freedom walking-assist exoskeleton has recently been developed for treadmill-based locomotion exercise for persons with neurological or orthopedic impairments. The TiltCyle system reported here uses two servo-motors to provide programmable assistance and resistance torques to the cranks of the two pedals.

1.2 History of the TiltCycle

Soon after its founding in 1979, the Palo Alto VA Rehabilitation R&D Center (RRDC) initiated a line of research in human lower extremity biomechanics and neuromuscular control. The focus of this research has been the understanding of functional use of the legs in locomotion [9]. Studies with both able-bodied subjects and persons with hemiplegia following a stroke were implemented to advance the understanding of normal and impaired gait and to improve lower-extremity therapy strategies aimed at re-establishing functional gait [e.g., 10, 11, 12, 13]. Central to this research has been the assertion that cycling is similar to ambulation from a neuromuscular control point of view, so that there would be both scientific carryover of knowledge to walking as well as therapeutic carryover for people treated using a cycling rehabilitation program.

The first TiltCycle, named for its ability to allow riders to pedal at any body tilt-angle from supine to upright [14], provided resistance through a Monark flywheel and circumferential band brake, and was used in the RRDC's Motion Physiology Laboratory through 1995. Periodic transformations ultimately included the replacement of the passive flywheel and band brake with two computer-controlled DC brushless servomotors that allow assistive as well as resistive pedal loads to be presented to test subjects. This paper describes these TiltCycle enhancements, the motor configuration, and control algorithms.

1.3 General Design Considerations

An ergometer is an exercise device that permits the experimenter to measure workload. This fundamental capability is dependent on sensors such as loadcells to measure the system's dynamics, and position sensors such as optical encoders to measure kinematic parameters in real time.

A second requirement is the ability to measure changes in human anatomy and physiology. Typical measures on a bicycle ergometer include goniometry of the leg joints, pedal forces, surface EMGs of leg muscle activity and exercise physiology data such as VO_2 and heart rate.

A third requirement is to present different stimuli to the subject. The simplest change could be in the workload itself by modification of the resistance torque. More complex interventions – in pedaling – are, for example, having one leg pedal forward while the other one pedals backwards (using kinematic coupling), or having each leg work at a different cadence, e.g., the left leg pedaling at twice the rate as the right. This paper will describe some of the advantages that computer-controlled pedaling resistance can offer the experimenter working with the TiltCycle ergometer.

2. Design of the Servomotor System

The TiltCycle, a clinical experimental device, has evolved considerably over 20 years of experimentation. The description below is of its current form.

2.1 Layout and Frame Description

The base is a 5 cm (2")-thick wooden slab 240 cm x 70 cm (96" x 28") on top of 14 cm (5.5") high longitudinal wood stiffeners and 6 height-adjustable legs for stability. The entire ergometer weighs approximately 240 kg (500 lb), with the two motors accounting for approximately half the mass.

The frame is made of Ø 32 cm (1.25") welded steel tubing with handlebar and crank bottom-bracket attachment points in a triangular arrangement (see figure 1). The base of this main triangle carries a longitudinal member to tie in the front and back feet and to hold the bearing races for drivetrain components. The crank bottom bracket, 43 cm (17") above the TiltCycle's base, also defines the hinge axis for the seat subframe (which carries a seatback as well) so that the seat-pedal distance is invariant with the seat subframe's tilt angle.

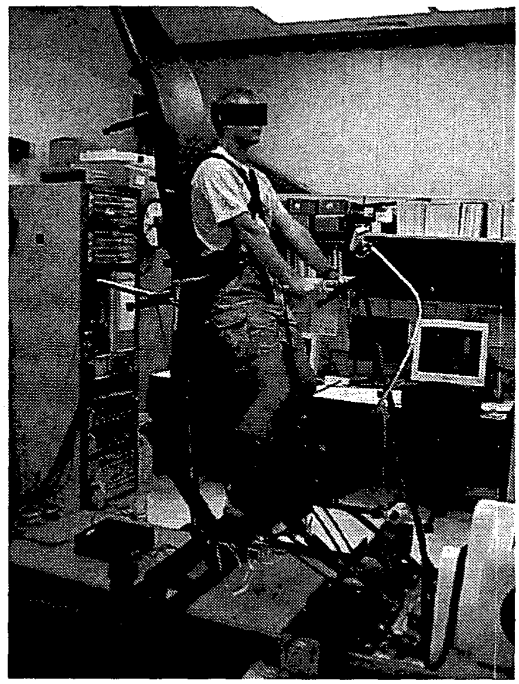

Figure 1: *TiltCycle showing 2 motors (bottom right), tilting frame and instrumentation rack (left), with the experimenter's console at right.*

The seat subframe carries a conventional bicycle seat and post as well as a padded, full-length seatback and straps. The shoulder and hip straps hold the cyclist safely onto the seatback; they also reduce postural sway and hip motion effects during cycling. The subframe tilt angle adjustment is powered, using a, non-back-drivable linear electric actuator that allows 0° (supine) to any angle up to 72° (conventional seat angle) adjustment in less than 10 sec by the operator while a subject is on the backboard. The hinges of the subframe are set 30 cm (12") to each side of the crank to allow ample room for the pedals and to allow a videocamera-based motion-analysis system an unobstructed view of reflective markers placed on a cyclist's shoes. This is used in studies that require real time measurement of ankle flexion/extension and motions out of the longitudinal plane during cycling.

The handlebars are mounted to the main frame with two over-center cam locks to permit easy vertical and longitudinal adjustment. The handlebar also carries a large stop button for the cyclist to trigger the E-stop circuit at any time and remove power to the motors.

2.2 Drivetrain

Two 10 KW (8.8 hp) brushless DC servo motors (Kollmorgen B606A, Radford, VA, USA) are mounted transversely on the wooden base forward of the cyclist (figure 2). Each motor shaft carries an adjustable torque-overload clutch [Morse Torq/Gard TG60, Ithaca, NY, USA]. The Kevlar-reinforced toothed-belt 4:1 ratio drivetrain is backdrivable, with a friction torque of 2 Nm measured at the crank, which is approximately 1% of the rated crank torque.

The pedal crank is split, unlike a conventional bicycle crank, such that each motor drives one pedal crank. The conventional crank with its pair of bearings was replaced with two half-cranks, each supported in the tube of the bottom bracket by two, close-set precision ball bearings. This arrangement provided for very low backlash and slightly more crank flex than the solid-crank setup, but the flex is not noticeable by subjects.

The maximal continuous (peak rating in parentheses) torque rating of the motor is 40 (200) Nm, producing 160 (800) Nm, or 120 (600) ft lb, at each pedal crank. The conventional 17 cm pedal crank arms allow each motor to produce a maximal tangential pedal force of 940 (4700) N against a cyclist's resistance. The peak torque rating can only be delivered for short, sub-second bursts and is limited in duration by the thermal time constant of the motor.

Each pedal carries a conventional SPD shoe cleat (Shimano Inc., CA, USA). The clip is screwed to a mounting block atop a 6-axis force/torque sensor (Model Delta-330, ATI, Garner, NC, USA).

2.3 Sensors

The host computer is a real-time data acquisition system, collecting data at a 1-kHz rate from:

- 16 surface EMG channels, typically 8 per leg,
- 12 pedal force channels: 3 force and 3 torque signals per pedal sensor,
- 2 motor tachometers, with a resolution of 14 RPM/V,
- 10 leg-joint angle goniometer channels (2 at each ankle and hip, one at each knee), and
- 4 4096-count/revolution optical encoders (2 crank and 2 pedal angle signals).

2.4 Data Acquisition

An Apple (Cupertino, CA, USA) Macintosh PowerPC uses two National Instruments (Austin, TX, USA) PCI-6071e multifunction data acquisition cards. Each card has 32 differential analog-digital converter channels (ADCs), 2 counters for optical encoder inputs and two digital-analog converters (DACs) for communication of a desired motor torque signal to the Kollmorgen motor controllers.

2.5 Motor Control

Two Kollmorgen D-20 controllers drive the motors. The conversion factor from voltage to crank torque is 266 Nm/V. The 12-bit DAC delivers an output resolution of 1.3 Nm per digital count.

The motors were chosen so that a 70 kg cyclist could produce approximately 1.5 bodyweights of force on each pedal continuously (8 bodyweights peak), which is beyond the capability of even elite cyclists.

2.6 Pedaling Control Modes

The computer-controlled ergometer must be able to provide an experience to the cyclists similar to that of a stationary exercise bicycle. The high-level real-time controller, which runs at a 1 kHz rate, calculates the desired motor torque T_m for each motor from the features that constitute the virtual system plant:

- a constant resistance torque similar to a band brake $T_c = K_c$.
- a virtual inertia torque proportional to angular acceleration, similar to the inertia of a real flywheel, $T_i = K_i\ \alpha$, implemented in digital form with an estimate of α derived from the tachometer signal: virtual torques of up to 100 times the plant's real inertia of 0.28 kg m^2 can be simulated (see section 3.1).
- a viscous torque similar to that experienced on some exercise devices that have air-resistance fans rather than flywheels with band brakes: $T_b = K_b\ \omega$
- a freewheeling safety feature similar to the ratchet mechanism on a bicycle that prevents the legs from being pulled around by the inertial load when the cyclist stops pedaling (if $\omega \leq 0$, $T_m = 0$).
- A lookup-table template torque profile ($T_t = K_t\ f(\theta)$, $0 \leq \theta < 360°$) that is used to provide an additional cyclic torque to the cyclist; it is used, for example, in one-legged pedaling to provide input from a virtual contralateral leg. The angular resolution of the table is 4096 counts per revolution.
- A PID (proportional, integral, derivative) closed-loop controller

$$T_{PID} = K_p\,(\theta_{des} - \theta_{real}) + K_v\,(\omega_{des} - \omega_{real}) + K_i\,\Sigma(\theta_{des} - \theta_{real}) \quad (1)$$

that cannot be used in conjunction with the above features, but is used when it is necessary to move a crank at a prescribed rate or to slave the motion of one crank with that of the other.

These single-leg, low-level system plant functions (except for the last one) are used in linear combinations to derive the desired torques for each motor. They are used by the following mid-level features that involve the coordination of both motors.

The kinematic coupling module connects the two cranks in software. In its simplest form, it allows the cyclist to pedal against a normal, bicycle-style, 2-legged load with pedals maintained at the conventional phasing of 180°. The kinematic coupling can be varied, such that one leg pedals at a different desired rate from the other, in either the same or the opposite direction; for example if the left is slaved to the right:

$$\omega_{\text{left-desired}} = K_{\omega}\, \omega_{\text{right-real}} \,. \tag{2}$$

In this case, the initial condition θ_0 is needed to fully describe the kinematics. In addition, the desired angular offset between the left and right sides can be set to an arbitrary amount, rather than the conventional $\Delta\theta = 180°$:

$$\theta_{\text{left-desired}} = \theta_{\text{right-real}} + \Delta\theta. \tag{3}$$

One motor is designated as "dominant" for the purpose of this algorithm; however, which leg this is has insignificant influence on the actual torque output. For the following example, let us designate the right leg as dominant. According to the conventional system plant model for resistance torque:

$$T_{\text{m-right-resist}} = T_c + T_i + T_b + T_t \tag{4}$$

The two cranks have kinematic states of $(\theta, \omega, \alpha)_{\text{right-real}}$ and $(\theta, \omega, \alpha)_{\text{left-real}}$. Based on the desired kinematic coupling, $\theta_{\text{left-desired}}$ and $\omega_{\text{left-desired}}$ can be determined. The torque to the left-side motor is now calculated from the PID control law:

$$\begin{aligned} T_{\text{m-left}} = {} & K_p\,(\theta_{des} - \theta_{real})_{left} \\ & + K_v\,(\omega_{des} - \omega_{real})_{left} \\ & + K_i\,\Sigma(\theta_{des} - \theta_{real})_{left} \end{aligned} \tag{5}$$

If the calculation were stopped here, the right leg would be pedaling against the conventional resistance torque, and the left crank would be kinematically slaved to the right crank. Any torque generated by the left leg would just influence the error terms of the left-motor PID controller, but would not contribute toward an acceleration of the virtual plant, governed by $T_{\text{m-right-resist}}$. To allow the left leg to contribute, the motor torque on the right side is therefore reduced by the left (non-dominant) side torque:

$$T_{\text{m-right}} = T_{\text{m-right-resist}} - K_k\, T_{\text{m-left}} \tag{6}$$

If $K_k = 0$, then the non-dominant leg is kinematically coupled to the dominant side. This is termed "pseudo-pedaling", since the non-dominant leg is not participating in the task of pedaling against the plant. If $K_k = 1$, then the cyclist feels as if both legs are contributing equally to the pedaling task. If $K_k = 0.5$, then the left leg has to pedal with twice the force as the dominant right leg to achieve an equal participation in the task against the virtual plant.

Another mid-level module allows for pseudo-pedaling, one-leg pedaling and the use of the torque template file. In the previous example, if $K_k = 0$, then the dominant leg does not feel the pedaling input of the non-dominant side, essentially performing one-legged pedaling against a system plant. This is very unnatural, since the one leg not only has to provide force in the downward power stroke (extension phase), but also has to pull up in the non-power portion of the cycle (flexion phase). To provide a virtual contralateral leg, the following procedure is implemented: during normal steady-state, two-legged pedaling at a particular cadence, the pedal forces are measured for 30 sec and then stored in a file. An average leg torque profile is calculated from 10-20 pedaling cycles for one leg. This "template" file $T_t = f(\theta)$ is added into the real-time controller on the dominant side, with pseudo-pedaling mode ($K_k = 0$) implemented. In steady-state mode at the same cadence, the cyclist now perceives with the dominant side that a virtual contralateral leg is assisting in the upstroke and adding some additional resistance in the power stroke, similar to a real contralateral leg. The non-dominant leg is pedaling normally, but, with $K_k = 0$, is pedaling only against the PID controller, and adding no real input to moving the virtual plant.

2.7 Experimenter Interface

The TiltCycle's experimenter interface is implemented in LabVIEW (National Instruments, Austin, TX, USA). The primary Virtual Instrument (VI) allows control of the Tiltcycle and the viewing of the data after each test run (Figure 2). This VI calls a lower-level VI to communicate with our LabVIEW Code Interface Node (CIN), which sets up a MacOS-managed interrupt-service routine (ISR) that runs the motor control code. The control code runs in 250 µsec each 1 msec with approximately 10 µsec variability in latency. The remaining 75% time is available for non-real-time, MacOS and LabVIEW code use.

The LabVIEW VI is the user interface for the experimenter. In addition to the main functions of starting and stopping the run, this VI handles parameter passing (e.g., changing the virtual inertia, modifying the length of the test run) and data collection from the log buffer for file storage. The CIN handles requests from the LabVIEW VI, passing parameters to the ISR via the state table on the right. The ISR is triggered every 1 msec by the data acquisition board interrupt, generated when a fresh set of 32 analog signals have been converted. At this point these signals are read, the

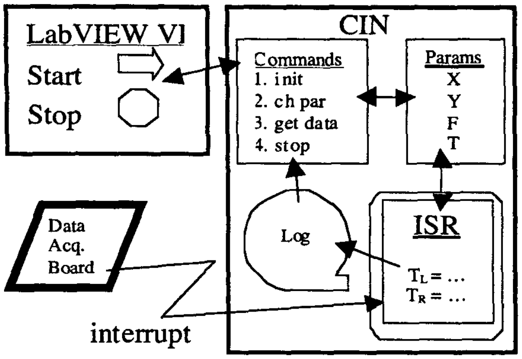

Figure 2: Overview of Software Design

encoder values are read, and finally the torque calculations are performed with the results output via the DAC ports to the two motor amplifiers.

3. Results of Validation Testing

3.1 System Identification

Using a least-squares estimation approach and a plant model as depicted in Figure 3 (see also equation 7), system identification parameters were derived. A number of experimental runs were done with multiple step inputs of widely varying torque levels.

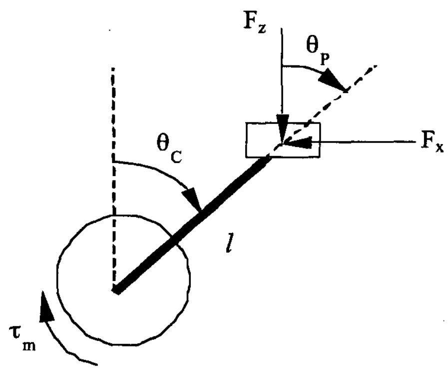

Figure 3: System model of the TiltCycle

$$I\theta_C = \tau_m + \tau_p + mgl\sin(\theta_C) - d\theta_C - \tau_f \qquad (7)$$

where:

I = crank intertia
τ_m = motor torque
τ_p = pedal torque, calculated from F_x and F_z
m = pedal mass
g = gravity
l = crank length
d = viscous damping
τ_f = friction torque

The application of the least-squares method resulted in the following estimated parameters:

$$I = 0.28\,kg\,m^2$$
$$d = 0.30\,Nms/rad$$
$$mgl = 2.75Nm$$
$$\tau_f = 2.0Nm$$

The damping, pedal mass and friction torques are all minor with respect to the pedaling torque levels typically applied by subjects, which can range up to 50 Nm, and have imperceptible operational effect with the implemented PID controller.

3.2 System Rigidity

Drivetrain rigidity was determined by applying a motor torque and constraining the pedals from turning using a fixed support. The negligible flexibility in the Kevlar toothed belt, preloaded according to factory specifications, the bearing flex at the driven-pulley half-crank, and the flex of the crank arm itself contributed to an error of less than one encoder count (<0.09°) at a loading torque of 50 Nm at the crank. Frame flex was rendered negligible by a pair of stiffener bars mounted between each of the motor plates and the crank bottom bracket. However, the PID control, using the 1-kHz control loop and a 4096-count per revolution encoder, is able to offer an effective proportional constant K_p = 2000 Nm/rad. This is considerably less stiff than a rigid crank would have been (estimated at 5000 Nm/rad), and led to ±5° variations in crank angle between left and right sides during regular pedaling against a heavy resistance load. However, this disparity was in practice not noticeable by any of our subjects, even after it had been brought to their attention.

4. Discussion

The plant model currently employed is limited to a simple second-order inertial system. We have noticed that certain disturbances, such as the friction torque and the torque due to the mass of the pedals, in aggregate contribute errors in the range of ±5 Nm. While this is not a first-order effect for the controller at the force levels normally desired, the control algorithms could be made more faithful by explicit forward modeling of these terms.

The encoder resolution, at 4096 counts per revolution, is the main limiting element in increasing the PID stiffness. While this has not been an experimentally important issue to date, it is likely to become a problem in the future as more sophisticated stimulus designs are required.

The real-time software is currently implemented in C as an OS interrupt service routine managed through LabVIEW. As more operational modes are desired, the complexity of mode shifting and parameter updates will start to render the operation of the system more difficult by the experimenter. Steps are currently underway to restructure the software to address this future limitation.

5. Conclusion

The TiltCycle enhancements presented in this paper have made it possible to perform a series of studies in interlimb coordination and muscle activity, contributing to the understanding of fundamental aspects of human neuromotor control. Without the ability to present programmable loads to the feet, it would have been necessary to realize the different stimuli in hardware, a much more difficult task. This system, since it is a one-of-a-kind device, will continue to be developed as the scientific questions evolve, and as new areas of inquiry in the field of biomechanics are opened. The quality of the motions and force stimuli are of adequate fidelity and dynamic range to extract useful information and draw conclusions from test trials with subjects, both able-bodied and those with motor control impairments.

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