Effects of varying the magnitude of the torque pulse on frequency locking and on entrainment

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***Abstract*—Human gait is a rhythmic process, whose frequency and phase can change as it interacts with another oscillatory process. This phenomenon is known as entrainment and it has been shown, in previous studies, to be able to alter walking speed and frequency. This thesis introduces a new method to influence human gait, by incorporating oscillatory behaviour into the magnitude of torque pulses applied at the ankle. The central objective is to explore how dynamic pulse amplitude can accelerate gait entrainment and achieve frequency locking. The thesis centres around the development of a high-level controller that modulates the pulse torque applied during walking. The work is divided into two sections. The first section uses a sinusoidal function to vary pulse amplitude. Experiments were conducted on a treadmill using an ankle exoskeleton system worn by a human participant. Although frequency entrainment was not observed, insight was provided into how modulation of pulse strength could affect gait dynamics. The second section introduces an alternative design: replacing the sinusoid with a nonlinear hybrid oscillator. This results in a more adaptable system to the gait signal via a coupling term. Multiple coupling strategies were explored, such as diffusive, adaptive and phase difference. Simulations showed that entrainment can be achieved using diffusive coupling. These results suggest that such hybrid oscillator-based controllers are feasible for a gait entrainment device. While experiments showed no entrainment, a more robust system, as well as more participants involved in the tests, could potentially demonstrate successful frequency locking and entrainment.**

I. Introduction

Human gait can be thought of as a series of actions starting from the initial impact of one foot and ending just before the next impact of the same foot. This sequence is named the gait cycle (seen in Figure [1]), and it repeats for as long as a person is ambulating. Furthermore, our understanding of the human gait can be expanded by describing it as a nonlinear limit cycle oscillator[1]. Once stabilized, the pattern of steps repeats over and over again, via self-sustaining oscillations. What distinguishes the limit cycle oscillator from other, simpler systems, is the way it interacts with external disturbances. When faced with a rhythmic perturbance, the oscillator will adapt its phase and frequency to match that of the perturbance[2][3]. As two oscillators start interracting, the gait and the rhythmic perturbation in the present case, they will begin to allign their phases, reaching a common frequency. The phenomenon present is named entrainment[4]. At first glance, it might seem as if the systems just simply synchronize, but that is not the case. Synchronization strictly refers to two events taking place at the same time. Entrainment is a more specific case of synchronization, that requires oscillatory behaviour from both signals[5][6]. This trait can be useful when studying neuromechanics of human gait, which could later aid in the development of gait rehabilitation therapy[7].

To prepare for the study of gait, it is necessary to create proper conditions for entrainment to occur. Traditionally, gait has been entrained via a torque pulse sent at a constant frequency[7]. One way of introducing the disturbance into the gait cycle is through a robot attached to the ankle. Its main function is to generate a periodic torque pulse during the gait. Previous studies have shown that both neurologically intact and impaired patients are capable of adjusting their cadence to match the period of repeated ankle perturbations, indicating that this approach to entrainment is effective [8][9]. Determining when to apply the torque pulse will first require a more thorough understanding of the gait cycle.

A singular gait cycle is called a stride. In most studies, the stride has been normalised from 0% to 100% in order to better describe the timing of events [4][10]. For a greater understanding, the sequence can be broken down into two main phases. Firstly, there is the Stance Phase, which describes the period in which the foot is on the ground. It can be seen in Figure [1] that the starting position is the heel strike (0%), which represents the initial contact of the foot with the surface. Since humans have different landing preferences on the foot's surface, from now on the initial or first contact terminology will be used. Loading response(0-9%) is the period in which the body weight is transferred to the foot to allow the other lower limb to move forward during the midstance(10-29%). During the terminal stage(30-49%) the heel rises again, which continues in the pre-swing(50-59%), where the foot is ready to be lifted off the ground again. Secondly, there is the Swing Phase, where as the name implies, the leg is in preparation for another heel strike. Starting from the toe-off(60-69%), the foot is fully lifted from the ground. Next, the leg is accelerated forward during the mid-swing(70-89%), only for it to be slowed down in the terminal swing(90-100%) part, right before the next contact with the surface[11]. With a complete map of a stride in mind, the application of the periodic torque pulse can be decided.

As a rule of thumb, the best way to incorporate a robot into a human's activities is for it to take away from the user's effort. Applying this mindset to the exoskeleton, the torque pulse should remove some of the effort brought by the movement of the legs, while not interrupting any step of the stride. We decided to send the pulse as the foot is being lifted off the ground. Placed precisely in the transition period between the pre-swing and the toe-off, the robot shares in the effort of the foot's movement. In previous studies, it was shown that as entrainment progresses, the user's stride duration adapts to match that of the perturbation. This has been observed through the convergence of gait timing over multiple strides [4]. Control for perturbation frequency can prove quite difficult, as the delivery of pulses is not consistent in all instances in the gait, potentially breaking the participant's gait rhythm and leading to tripping. For the scope of this research, it is desirable to have the pulse frequency at a fixed position in the sequence. From the literature explored, the perturbation was decided to be introduced at the 50% mark in the gait cycle. This alone will not produce entrainment, as the walking process is only one signal present in the system. The challenge then becomes to entrain the user through different means. Since entrainment works by having two oscillators influencing each other, the magnitude of the pulse was chosen to become the second oscillator. This leads to the central research question of this thesis: How will the application of a torque pulse with an oscillating magnitude, at a different frequency from the stride frequency and applied at a specific gait-phase, affect the entrainment of users, as well as achieve frequency locking.

The desired outcome of the ankle robot is to match the frequencies of the gait with the fixed perturbation frequency. In the previous papers [4], the convergence of the two has led to a preferred timing for the delivery of the torque pulse in the gait cycle to emerge, where the entrainment occurs. The consequence of the frequency matching, when the timing was periodic, was that the phases aligned themselves in what is called phase-locking. Consistent results in entrainment were shown when the perturbation was sent at approximately 50% of the sequence, where the stages involve a high level of mechanical work [4] [10]. While this was found to be an optimal point in the cycle, phase locking is still highly susceptible to the reflexes and mechanical preferences of the users, which can find better entrainment, for example, towards the beginning of the cycle, at first contact, present in one of the subjects in [9]. In all of the studies explored for this paper, a focus was placed on achieving phase-locking with a fixed magnitude for the perturbation. For the proposed research, it will be attempted to use oscillatory magnitude to achieve entrainment. With each stride, the magnitude would converge to an optimal value that matches the oscillation of the gait cycle. As a consequence the system would experience magnitude-locking as well as frequency-locking.

II. Methods

In order to generate entrainment, an exoskeleton attached to the user's leg is used. Originally designed to improve balance by canceling disturbances, the device seen in Figure[2], delivers a torque pulse, in the plantar-flexion direction, at the ankle[12]. Two sections are intertwined to make up the exoskeleton, the low-level controller and the high-level controller. A low-level controller represents the individual components that make up a system as well as its general functions, while a high-level controller is responsible for the application of these functions. Now to relate it to our topic, the low-level controller governs the torque generation of the exoskeleton, and how the motor runs the ball screw in order to tighten or loosen the Bowden cable. The motor's velocity is converted to torque this way, and is applied directly to the user's foot. Where the high-level controller comes into play is the application of torque. The timing, magnitude and shape of the torque signal is all decided by the high-level design. The research conducted by us is strictly focused on the application of torque in a specific manner, such that entrainment can be achieved.

In this section, the high-level controller used to deliver the pulses is explored. As a baseline for developing our system, the high-level controller from [7] was used. For their purposes, the pulse was delivered with a constant amplitude, where the phase varied in order to find the best spot in the gait phase to deliver the pulse. Now the point of delivery was chosen at push-off(50% of stance), and the magnitude of the pulse will be varied. The shape of the pulse is trapezoidal. To account for the small delay in the delivery of the pulse by the exoskeleton, as well as keeping the cable tight, a pretension torque is implemented. The best place to start the dissection of the torque pulse modulation is at the initialization step. In this part, most of the fundamental design decisions are made. We chose the base value of the torque as 10 Nm, a value carried over from the previous study [7], which was found to be safe for the participants. For our purposes it will serve as a midpoint during the modulation of the desired torque magnitude. The pulse is further described to have a duration of 150ms, with only 20ms duration at the peak. Tension is applied to the cable 50ms before the pulse delivery in order to avoid a delay in the desired timing. The application is further defined by a pulse phase variable, which dictates when it should occur during the gait phase. During the design process of the controller, it was set at 50% of the stance phase, during mid-stance to be precise. Further in the paper, experiments will be discussed in which the pulse was applied near the terminal stance(80% of the stance phase). Finally the pulse was decided to be applied within 3% of the phase of the intended timing.

Now that we have the shape of the pulse in mind, we have to determine under what conditions the user will entrain. Our approach starts with determining the preferred stride frequency of the person. In order to get insight into the gait of the user, we have used as input analogue signals from a split treadmill(isolating each leg). Important for the high-level controller are the ground contact signals, which determine via the use of force plates, whenever one of the feet are touching the ground, the gait phases of both legs(stride gait phase, percentage of stance and of swing) as well as the gait durations(stride, stance and swing). Since frequency is inversly proportional to time period, an estimator block was made to calculate the preferred stride period of the user. Each stride is

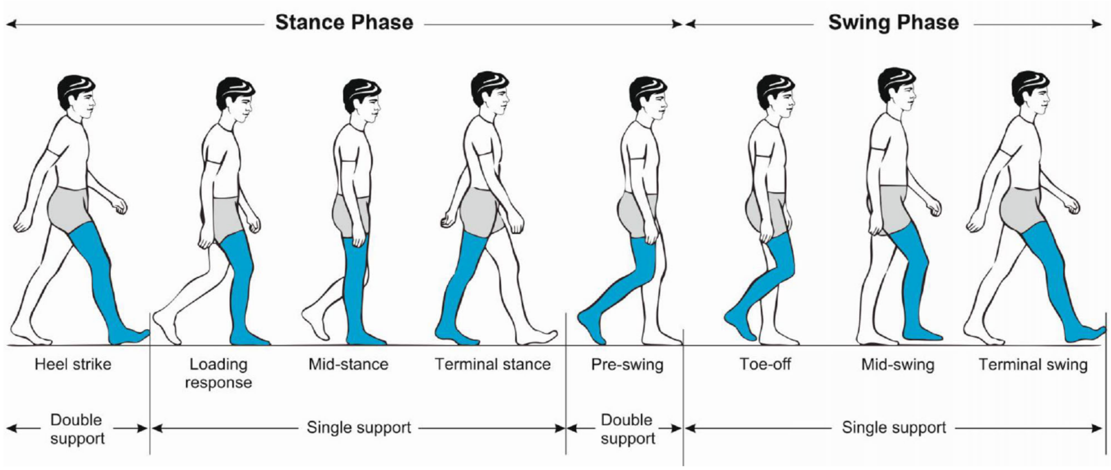


Fig. 1: Full visualization of the gait cycle of a human body[11].

recorded inside a stride counter function, based on the phase of the user in the gait cycle. If the current value is smaller than the last recorded phase, then a new step was taken. With the use of a simple clock signal, the time between each change in steps is recorded. One risk of a constantly changing period is forcing the user to start speeding up, in their effort to try and match the pulse. To avoid this, we calculate the mean of the stride period over ten strides. As was stated in the previous section, in order to complete a stride, the user has to finish on the same leg that started the tracking. Both the counting of the steps, as well as the average stride period are calculated for each leg individually. An enable input is also included, to recalibrate the stride period whenever necessary.

With the collection of the gait data and the shape of the pulse decided upon, the oscillator can be constructed. The final block of the high-level controller wraps it all together with the generation of torque pulses. The shape of the torque pulse is given by the equation 1, with the amplitude determined in equation 2.

$$\tau_{\text{exo}} = \begin{cases} \tau_{\text{PT}} + \dfrac{\tau_{\text{mod}}(t - t_{\text{PO}})}{40}, & 0 \leq (t - t_{\text{PO}}) \leq 40 \\ \tau_{\text{mod}}, & 40 < (t - t_{\text{PO}}) \leq 60 \\ \dfrac{\tau_{\text{mod}}(t - t_{\text{PO}} - 100)}{40}, & 60 < (t - t_{\text{PO}}) \leq 100 \\ 0, & \text{otherwise.} \end{cases} \quad (1)$$

$$\tau_{\text{mod}} = \tau_{center} - A * sin(2\pi f \cdot C) \quad (2)$$

The τ_{PT} is the pretension torque, the t_{PO} is the time of the start of the pulse, and the τ_{mod} is the torque modulated with a sinusoidal signal. The coefficient C gives a small offset to the frequency of the exoskeleton, and the constant A represents the amplitude of the sinusoidal wave. This, together with the central torque point τ_{center} determine the minimum and maximum magnitude of the pulses. It is necessary for the difference between the frequency of the oscillator and the gait frequency to be small enough(at a maximum of 5%) for the user to passively be influenced by the stimuli. The sine wave runs continuously in the background, and the delivery of pulses is based on the current gait phase and the desired timing within the stance phase. Within its structure, unless the gait phase is at the specified phase in the stance, the output will always tell the generator to send the pulse at the 200% mark of the stance phase. Such a thing is impossible, and as such, the torque output is zero. This logic helps prevent interference from overly frequent or poorly timed pulses. The function evaluates gait phase input (gait phase) and compares it against a desired pulse phase. Based on the relative timing of these phases, as well as conditions like swing or stance state, the function determines when a pulse should be initiated. Outputs include the desired torque for each leg (torque exo desired).

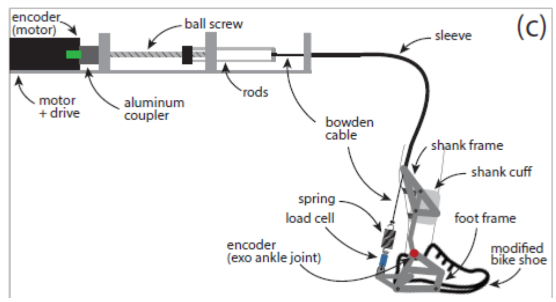


Fig. 2: Diagram of ankle exoskeleton[12]

A. Theory of Nonlinear Oscillators

While magnitude modulation as a concept shows great promise to reach a state of entrainment, we have reasons to believe there are ways in which it could be improved. Looking back on the interaction between the user's gait and the torque pulse, once the ankle is stimulated, the gait slowly converges to the frequency of the exoskeleton oscillator. A more faithful realisation of entrainment would have both participants change in order to reach an equilibrium point. In the previous section, we saw that the delivery of a torque pulse with a sinusoidal pattern is indeed possible. Going forward we want to keep this behaviour, while having the gait directly influence the torque pulses. By having the oscillators affect each other, our belief is that the convergence to the desired frequency would be more robust. In order to construct the new torque pulse, we have looked back on previous studies that have investigated gait. By modelling the signal that gives the magnitude based on limit-cycle oscillators, it would have in its composition a coupling term. This way, the gait can directly influence the torque pulses.

1) Van der Pol: The most well-known example of a limit-cycle oscillator is the Van der Pol oscillator. This nonlinear dynamical system is capable of exhibiting self-sustained oscillations. It was originally developed by Dutch physicist Balthasar Van der Pol while studying electrical circuits containing vacuum tubes. The system itself can be thought of as an RLC circuit(an electrical system consisting of a resistor, inductor and a capacitor), which oscillates by exchanging energy between the capacitor and the inductor. Governing the system are the differential equations 3 and 4, where x is the state variable representing the capacitor voltage and y representing the inductor current. What sets this system apart is the resistive component, switched to act as a nonlinear damping term μ. This element has a proportional relationship to the voltage of the system, exhibiting negative damping when the energy of the system is low, and switching to positive damping when the system reaches high levels of energy. A push-pull relationship is created, amplifying low amplitude oscillations($|x| < 1$) and attenuating high amplitude oscillations ($|x| > 1$), resulting in a limit cycle oscillator[13].

$$\dot{x} = y \tag{3}$$
$$\dot{y} = \mu(1 - x^2)y - x \tag{4}$$

What prevented us from directly using this system is the lack of tunability of the frequency. In the following section, we will discuss how the system can be altered to not only include a method to control the frequency of the oscillator but also a more robust nonlinear damping component.

2) Rayleigh: Another variation of a limit cycle nonlinear oscillator is the Rayleigh oscillator. Originally developed in ”The Theory of Sound” paper[14], it differentiates itself from the Van der Pol oscillator by modelling the nonlinear damping term only using the time derivative of x, as it can be seen in equation 6[13][1][3]. This second method of generating a self-sustained oscillator has been used in previous studies to model gait behaviour[15]. This was done by having x be the position signal and y as the velocity.

$$\dot{x} = y \tag{5}$$
$$\dot{y} = -\mu(y^2 - 1)y - x \tag{6}$$

3) Hybrid (needs to be better incorporated): To model the complex dynamics of human gait, especially the lateral pedestrian force that arises when a person walks on a moving surface or interacts with an exoskeleton, researchers have proposed a modified version of the classic nonlinear oscillator. The lateral pedestrian force refers to the oscillatory side-to-side load applied by a person to the ground while ambulating[16]. The proposed formulation enables the generation of both self-sustained oscillations as well as frequency entrainment, two points essential to replicating human gait. Another point to mention is that coupling multiple hybrid nonlinear oscillators, each representing distinct joint motions (such as hip, knee, and ankle), has been used to create networks capable of coordinating rhythmic movements across multiple degrees of freedom [1]. This plays into our effort to establish an oscillator that can closely mirror the gait control system inside a human body.

The hybrid oscillator discussed and implemented in this paper is a nonlinear oscillator that combines elements of the Van der Pol and Rayleigh oscillators with an additional frequency control term. Seen in equation 8, this is an adapted nonlinear oscillator from the paper[15]. Now, to dissect the contents of this new system. First, we have a nonlinear damping term. The parameter ϵ is based on position, allowing the system to have self-sustained oscillations by amplifying low-energy and damping high-energy deviations. Secondly, γ acts as a bias for this damping term, affecting how the oscillator's amplitude grows or decays. The parameter p sharpens the nonlinearity in the position-based damping, making the system more robust to deviations. The offset x_0 horizontally shifts the oscillator's centre, effectively biasing the output signal to oscillate around a specific torque value. For the use case of this system, it was set to zero, since we already have a term that acts as a baseline torque. Moving further, the term $\delta(1-qy^2)y$ introduces damping based on velocity. This represents the Rayleigh-inspired section of the oscillator, where δ is the scaling factor of the damping and q adjusts the suppression of high velocity amplitude. Finally, ω is the natural frequency, allowing us to directly tune the frequency of the oscillator[1].

$$\dot{x} = y \tag{7}$$
$$\dot{y} = \epsilon(\gamma - p(x - x_0)^2)y + \delta(1 - qy)^2)y - \omega^2(x - x_0) \tag{8}$$

4) Coupling methods : Now with a more robust oscillator, capable of finer frequency tuning and closely replicating gait behaviour, we can delve into the second point of interest of this system. The coupling of two signals can be thought of as a feedback network established between them. Changes in one will result in alterations of possibly the shape, amplitude and frequency of the other. Looking at already existing literature, coupling can be split into two categories, attractive and repulsive. As the name implies, attractive coupling helps bring together two different signals by reducing their differences.[17]. To bring together the gait and the hybrid oscillator, we have looked at two characteristics that can be easily changed in both, from instance to instance.

The first thought that came to mind was to use the phase of both signals. This coupling strategy would rely on the alignment of the phase of the nonlinear oscillator and the gait cycle phase, by differentiating both and adding the term to the derivative of the position term in the hybrid oscillator, as seen in equation 9. The K term governs the strength of the coupling. This strategy is inspired by synchronisation approaches seen in neural oscillators and adaptive frequency systems[18][19]. While this method can be advantageous by promoting stable phase alignment, it also introduces sensitivity to phase estimation noise. The influence the phase has on the nonlinear oscillator will be discussed further in the simulation section.

$$\dot{x} = y + K \cdot (\phi_{\text{gait}} - \phi_{\text{osc}}), \tag{9}$$

The second proposed method is diffusive coupling, which operates strictly with the instantaneous difference between the amplitudes of the two oscillators. The coupling term is expressed in equation 10, where x_{gait} is the real-time gait measurement (for example stride-normalized position or estimated torque) and x_{osc} is the amplitude of the oscillator output. Same as before, the K term represents the strength of the coupling and how much it affects the oscillator.

$$\dot{x} = y + K \cdot (x_{\text{gait}} - x_{\text{osc}}) \tag{10}$$

Diffusive coupling takes advantage of simplicity and fast convergence, as it does not require any unwrapping of the phase. The inspiration for this coupling method came from similar diffusive approaches that have been successfully applied in studies of coupled limit-cycle oscillators[1]. One risk of this method, which will be discussed shortly, is a possible phase offset if the coupling is not properly tuned.

III. Experimental Setup and Protocol

One participant(a 22-year-old male, weighing 95kg) was equipped with the exoskeleton on both the left and right lower legs. Only the exoskeleton worn on the left leg has applied short and small pulses of plantarflexion torque. He walked on a split treadmill at a speed of 0.8m/s. To mitigate any risk of injury caused by falling or tripping, the participant wore a safety harness connected to the ceiling.

The experiment consisted of two sessions, one where the functionality of the high-level controller was tested and a second one where different configurations of the controller were tested in short sequences. The first session served as an accommodation period for the exoskeleton, where we could verify if the simulated torque pulses, seen in Figure [3], could be replicated by the robot. Vital to our study was the discovery of any discomfort present in the torque delivery, which could negatively impact the experiment. After observing the successful application of the pulse, several adjustments were made. At first, the range of the pulses varied between 5Nm and 15Nm. Consequently, the range was changed to 7Nm to 13Nm towards the end. Following this alteration, the difference in frequency between the two oscillators was changed from -5%(seen in Figure[3a]) to -2%(seen in Figure[3b]). Lastly, two different gait phase timings were also explored, one being at 50% of the stance phase, while the other was at 80% of the stance phase.

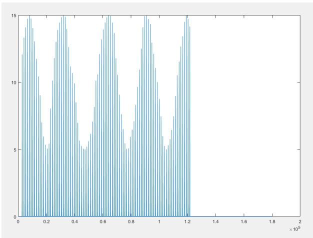
Returning to the laboratory for the second round of experiments, a structure was required for a better collection of data. Changes in the frequency of the sine as well as the phase timings were picked to form the protocol of the experiment. A full list of all the conditions is found in Table[1]. The amplitude of the pulses remained in the range of [7Nm, 13Nm]. Before we could begin, the calibration of the treadmill, as well as of the ankle exoskeleton, was performed. After the user was fitted with the exoskeleton, he was asked to start moving on the treadmill at a relaxed pace. After a few strides, the average gait frequency was computed. The experiment itself consisted of a series of 8 walks, each having torque pulses delivered to one of the feet for exactly three minutes. After the completion of a single walk, the user gets 45 seconds of rest, with no disturbances applied to their gait. A new stride frequency is then calculated in 15 seconds, and a new walk with different experimental conditions is started. Inside the Simulink file that contains the high-level controller, a protocol function was implemented, which returns the 8 pairs of conditions in a random order, thus stopping the user from preparing unconsciously before the start of the pulse delivery. This process is repeated three times in total, to account for uncertainties brought upon the ordering.

Frequency Oscillator	Stance Phase Timing
$95\% \cdot f_{gait}$	50%
$98\% \cdot f_{gait}$	50%
$102\% \cdot f_{gait}$	50%
$105\% \cdot f_{gait}$	50%
$95\% \cdot f_{gait}$	80%
$98\% \cdot f_{gait}$	80%
$102\% \cdot f_{gait}$	80%
$105\% \cdot f_{gait}$	80%

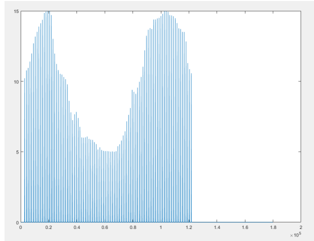
TABLE I: Frequency oscillator and stance phase timing data

IV. Results

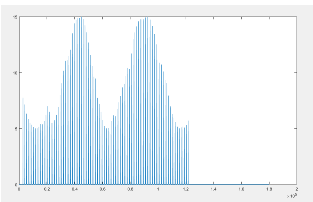
Circling back to the theoretical area for just a brief moment, before the experiment was realised, the high-level controller had to be checked. A simulation environment was realised inside Simulink for these purposes. To verify the torque output of the high-level controller, a recording of 2 minutes of gait data, gathered from the treadmill, was used as the input. Four different frequencies were used to generate the waveforms seen in Figure[3]. We chose a maximum of 5% difference in frequency, as not to stray from the orbit where entrainment is still achieveable. The shape formed by the pulses can be seen to be briefly distorted at the beginning, until they stabilise and form a coherent sinusoidal pattern. It is also noticeable that the shape of the sine wave becomes more ample and well-defined as we get closer to the actual frequency of the gait. This is in line with how the pulses are expected to manifest, as the oscillator would gradually slow down.



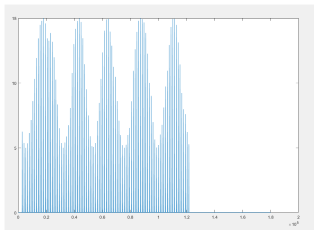
(a) Torque pulses at 95% of gait frequency



(b) Torque pulses at 98% of gait frequency



(c) Torque pulses at 102% of gait frequency



(d) Torque pulses at 105% of gait frequency

Fig. 3: Torque pulses at different percentages of the gait frequency

On the other side of the thesis, coupling simulations have been conducted. Inside Simulink, both of the coupling methods have been tested in entrainment trials at first with other hybrid oscillators of the same kind. The test was done with a 2% difference in frequency, and it resulted in an extremely fast entrainment for the diffusive coupling method, while it only showed a synchronisation in the phase at a fixed interval for the phase coupling method, as well as distortion of the shape of the signal. Moving forward from a coupling of two near identical signals, a sine signal has been constructed using a frequency determined with ground contact data from the treadmill. The resulting frequency was equal to 0.783Hz. Both the phase difference and the instantaneous amplitude difference couplings were coupled to this signal, as can be seen in Figure[4]. While the diffusive coupling entrains to the signal extremely fast, the same cannot be said for the phase coupling. That signal has a frequency of 0.693Hz, a more than 10% difference between the oscillator and gait. Another attempt was made to use a recording of the gait phase inside the coupling term, but it resulted in no output coming out of the hybrid oscillator, even at a very low coupling strength(K = 0.01).

In closing the results section, we will present the data gathered from the experiment. Out of the proposed trials with 8 different conditions, only three conditions were able to be tested. This came as a result of both issues with the calibration of the ankle exoskeleton and with the implementation of the protocol delivery function, which severely cut the time of the experiments short. Presented in Figure [5] are the torque pulses sent to the left foot of the user. The first walk has been conducted with the torque delivered at 50% of the stance phase, while the other two have been delivered at 80% of the stance phase. As for the frequency percentage at which they were delivered, in order, they are 98%, 102% and finally 105%. The only section that shows the potential of entrainment is the last performed walk, in the valley of the torque, due to repeated pulses of the same amplitude. While this area could be considered too small for any meaningful entrainment to be present, it can still show signs of a maintained magnitude. The phase of the sinusoidal signal was plotted at the area of apparent entrainment(between 5200ms and 7000ms), and compared to the mean of all the calculated phases in this interval. If the phase values oscillate close to the mean value, then the pulses would be consistently delivered at the same magnitude. As it can be observed in Figure6, this was not the case. With the mean phase equal to -0.087 rad, it is clear that all of the torques differ too much from the calculated average.

V. Discussion

A. Experiment

One key takeaway from the experiment is the lack of entrainment present during the walks. There are two main observable criteria that can validate if the user is entrained. First, we can look at the magnitude and whether it stays constant for multiple pulses in a row. While this was achieved for brief periods of time, only for the peaks and the valleys of the signal, it is not sufficient to confirm entrainment. The second indicator is for the phase of the gait to remain the same across a series of strides. As stated earlier in the paper, the phases diverge too much from the mean of the phases to be consistent.

Taking a look back at the experiment, we can discuss the manner in which it was conducted, and how that later feeds into our measurement of frequency locking and entrainment. To start, the user has worn only one exoskeleton, on the left leg. This came as a result of an inability to perform a proper calibration of the right leg exoskeleton. Such an imbalance between the legs has direct consequences on the walk of the user, from different forces applied to each step, to the mind being more focused on maintaining the balance, which would make it harder to entrain. The difficulties in the calibration have also resulted in less reliable delivery and tracking of torque pulses, which could have made the process of entrainment more difficult. During the experiment, the protocol function also tended to repeat past conditions, requiring manual resets each time. Another factor can be the individual. There is a precedent for select cases in which the application of pulses does not affect their gait. Using only a single user hindered the generalizability of the findings and increased the risk that observed changes in gait were because of voluntary behavioral adaptations. Another issue which aids to the theory of voluntary behavioral adaptations came in the form of the fitment of the device. Some discomfort has been felt while delivering the pulses, as the brace would push into the shin at higher torque values. This can, in turn, create a reflex to try to mitigate the effect felt on the shin, which would again disrupt gait. Lower magnitude peaks were generally better tolerated, and more favourable responses were observed when pulses were applied at 80% of the stance phase.

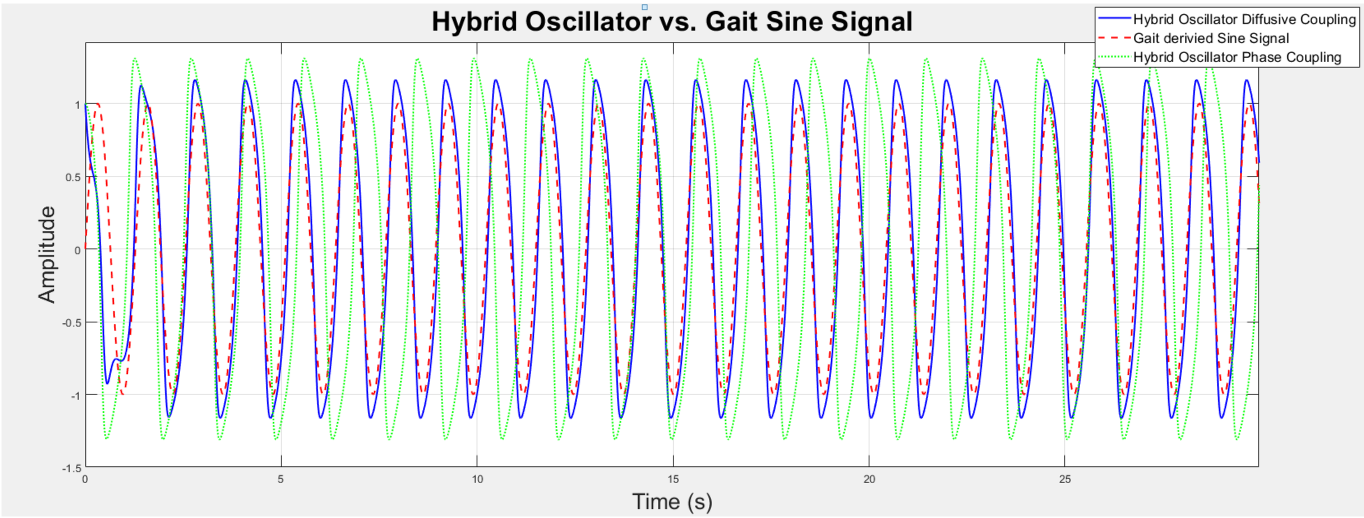


Fig. 4: Nonlinear hybrid oscillator coupled with a sinusoidal signal at gait frequency. Two methods of coupling have been used.

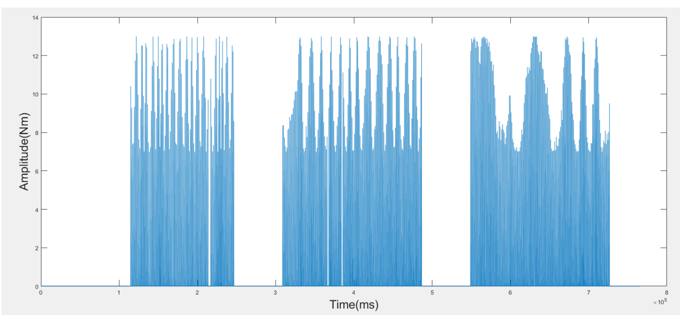


Fig. 5: Magnitude plot of the torque amplitude sinusoidal signal.

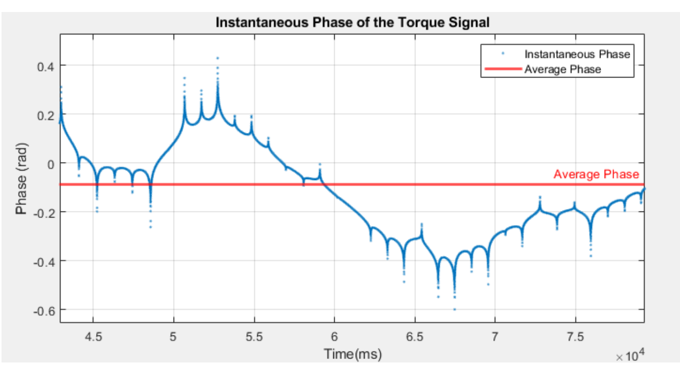


Fig. 6: Phase plot of the torque amplitude sinusoidal signal.

With all of these limitations in mind, a future experiment would have to be conducted with a larger number of participants. The proposed protocol would have to be followed thoroughly, with multiple walking sessions per condition to be certain of the collected data. The amplitude should also be decreased, most likely maxing out at 10Nm to maintain the user's comfort in mind. This way, possible adaptations to pain can be avoided.

B. Hybrid oscillator

For the second part of the thesis, only simulations of the hybrid nonlinear oscillator have been conducted. From the different implementations discussed in this thesis, it was observed that the hybrid has a very sensitive frequency control. The frequency of the oscillator is not only affected by the natural frequency term, but also by the tuning of the components that make up the system. This is because it is a limit cycle oscillator, and the self-sustained oscillations are being maintained by the dampening terms. As a result, a fine tuning of the oscillator would be crucial for a future implementation of the magnitude-modulated torque pulse delivery. A negative aspect of this behaviour has been seen when attempting to achieve coupling through phase difference. Trying with multiple coupling strengths, the difference was seen to be too great, even at the start. This would immediately destabilise the hybrid oscillator, which would either result in a higher phase difference and lead to a death spiral for the oscillator, or a severe shape deformation, together with a lowering of the frequency. Tests with the real recorded gait phase data highlighted this the most, as they would always lead to zero oscillations from the start. A possible solution to this problem would be the application of the coupling term with a time delay from the start, thus letting the oscillator stabilise before the feedback from the gait can be applied. This should also come paired with a very low coupling strength value.

As for the other coupling method, it showed great results in frequency locking and then entraining. Even with 5% differences in the frequencies of the two oscillators, the hybrid would entrain to the gait frequencies within 20 seconds. One factor that can have a negative effect on the signal is the coupling strength. With poor tuning, it will still lead to frequency locking, but a phase offset would be introduced as well. This could potentially harm entrainment when used in tandem with magnitude modulation, as the user could not accept the current pulse magnitude, even if the frequencies are close to alignment.

VI. Conclusion

This thesis set out to explore how modulating the magnitude of torque pulses, rather than just their timing, could influence the entrainment of human gait. This was realised using an exoskeleton system worn at the ankle. Rooted in the concept of gait as a nonlinear limit-cycle oscillator, this work reimagines the perturbation signal not as a rigidly timed stimulus but as a dynamic, interacting process capable of adapting in amplitude through coupling with the user's gait.

Initial experimental results did not demonstrate entrainment using traditional sinusoidal modulation. However, simulations of the nonlinear oscillator revealed a clear contrast: diffusive coupling, where the torque magnitude was adjusted based on the real-time difference between the gait signal's instantaneous amplitude and the oscillator's state's instantaneous amplitude, which in turn resulted in stable frequency locking and entrainment.

Although entrainment was not observed experimentally, the simulations validate the feasibility of the proposed hybrid controller structure and highlight its potential for future development. Limitations such as the single-participant setup, hardware constraints, and calibration issues likely contributed to the lack of observable entrainment in practice. These challenges only highlight the need for further experimental trials with improved mechanical design, smoother pulse transitions, and multi-user testing.

In summary, this work demonstrates that adaptive modulation of torque magnitude, especially via diffusive coupling, is a promising direction for human-exoskeleton coordination. Future efforts should focus on refining the hardware interface and validating these findings in broader populations to realize the full potential of gait-adaptive exoskeleton systems.

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APPENDIX

During the preparation of this work, the author used AI tools (Grammarly) to revise grammar and improve sentence flow. After using these tools, the author reviewed and edited the content as needed and takes full responsibility for the content of the work.