CHAPTER 7

EPILOGUE
The overarching aim of the work presented in this thesis was to assess and understand the effort for balance control in terms of the metabolic cost of walking in able-bodied people and patient populations. Where the previous chapters focused on specific questions that can contribute to this aim, this Chapter will reflect on the results in light of the overall aim. First, general conclusions regarding the energy cost for balance control in able-bodied people and patient populations will be made and discussed. This will be followed by a discussion of the potential sources of this cost based on the observed changes in step parameters and muscle activation. In section three, methodological issues pertaining to the experimental studies will be discussed, resulting in future research directions in section four. Finally, this chapter will end with clinical implications of the research performed in the course of this thesis.
THE ENERGY COST FOR BALANCE CONTROL

The energy cost for balance control in able-bodied people

Establishing the sources of the metabolic requirements of walking is imperative to understand and improve the decreased gait economy in patient populations. In this thesis we aimed to assess the energy cost for balance control to determine whether the decreased gait economy in patient populations is (partly) due to an elevated metabolic effort for balance control. First we aimed to establish this cost in able-bodied people to provide a reference. In Chapter 2 we used an external stabilization device to assess the energy cost for balance control while walking on a treadmill. Using a similar set-up but with arbitrarily chosen stiffness, previous studies had found reductions in energy cost ranging between 3.2-5%\textsuperscript{47-49}. By varying stiffness we were able to reduce the energy cost of walking up to a maximum of 7.5% (with a spring stiffness \( \geq 1260 \text{ N m}^{-1} \)). This reduction in energy cost with external stabilization relative to normal treadmill walking without stabilization can be regarded as a lower limit of the energy cost for balance control during walking.

Walking without disturbances as in Chapter 2 may not well reflect the energy cost for balance control in daily life, in which challenges to balance control occur due to disturbances and limitations with regard to the use of balance control strategies. We expected that such situations would result in an increased effort for balance control. Indeed, in Chapter 3, we observed increases in energy cost of 13% in the high postural threat condition. Similar increases in energy cost have been found in other studies where stepping strategies were constrained without the explicit aim of inducing postural threat\textsuperscript{18,54}.

The two studies in healthy subjects combined provide insight at the range of magnitude of the metabolic effort for balance control in able-bodied individuals. For the imposed conditions this ranges between 7.5-20.5% depending on the level of challenge to balance control. This might of course be even higher for more extreme balance manipulations. However, such manipulations will involve increasing amounts of external energy imposed on the participant. The dissipation of this externally imposed energy will dominate the metabolic cost independent of balance strategies used.
Chapter 7

The influence of walking speed on the metabolic effort for balance control

Whether or not slow walking is more stable has been an ongoing debate in the literature. We expected that, if slow or fast walking would be more stable, the metabolic effort for balance control would decrease accordingly at slow or fast speeds. To investigate this, the manipulations of balance control in the studies with able-bodied individuals in Chapters 2 and 3 were applied at different walking speeds expressed as percentages of the participants’ preferred walking speed, ranging from 70-130% for the stabilization study (Chapter 2) and from 60-140% for the postural threat study (Chapter 3). On average, the slowest speed (0.8±0.07 m·s⁻¹) was close to the speed deemed necessary for community ambulation ¹⁵⁹, while the fastest speed (1.8±0.17 m·s⁻¹) was close to the walk-run transition speed ¹⁶⁰. In both studies, walking speed did not have a significant effect on the energy cost for balance control, although the effect of postural threat tended to be largest at the most extreme speed conditions. These results show that the effort for balance control is independent of walking speed, and thereby contribute to the growing body of evidence refuting the notion that a slow or fast walking speed is more stable ⁵⁹,⁶¹.

The energy cost for balance control in lower limb amputees and stroke survivors

In Chapter 2-3 we investigated the energy cost for balance control in able-bodied individuals. These studies can serve as a reference for the studies in lower limb amputees and stroke survivors. These patient populations often suffer from an elevated metabolic demand for walking ¹-³. In this thesis the aim was to assess whether (part of) this increase may be explained by an elevated metabolic effort for balance control. Unfortunately, the study in Chapter 4 with lower limb amputees does not permit a comparison between the energy cost for balance control in this group and able-bodied people. In this study we used the same stabilization set-up as in Chapter 2. However, the external stabilization led to unexpected results in the lower limb amputees. Especially the transfemoral amputees appeared to be hindered by the springs, and showed an increase in energy cost when walking with external stabilization. Due to this apparent negative effect of the stabilization device, a sound
comparison of the energy cost for balance control in people with a lower limb amputation compared to able-bodied individuals cannot be made.

The studies in stroke survivors do allow such a comparison. With this group we used a cane and handrail to facilitate balance control. Besides benefits with regard to balance control, cane use also brings about a metabolic cost for holding and carrying a cane. In able-bodied individuals cane use has been found to increase the cost of walking up to 33% \(^{123,125}\). In Chapter 5 we found that in stroke survivors who are no longer dependent on a walking aid in daily life, cane use also increased the cost of walking, although this effect (6.1%) was far less than in able-bodied individuals. By contrast, the dependent walkers showed a decrease in energy cost of 8.4% when walking with a cane, indicating that stroke patients have larger benefits of cane use than able-bodied people.

The effects of a balance aid on the energy cost of walking were magnified on a treadmill when using a handrail instead of a cane, as evidenced in Chapter 5 (19% reduction; 1.36 J·kg\(^{-1}\)·m\(^{-1}\) for dependent ambulators) and Chapter 6 (11% reduction; 0.95 J·kg\(^{-1}\)·m\(^{-1}\)). This larger difference can be explained by the fact that no cost of holding or carrying a cane was involved, and potentially because walking on a treadmill is more challenging to stroke survivors than over-ground walking, as implicated by the higher energy cost during treadmill compared to over-ground walking.

The absolute reduction in energy cost with handrail support in stroke patients, 1.15 J·kg\(^{-1}\)·m\(^{-1}\) on average, can account for approximately 33% of the difference in energy cost of walking between the stroke survivors and the able-bodied people when walking at preferred speed. This shows that balance control plays a substantial and clinically relevant role in explaining the decreased economy in this patient population. The relative reduction in energy cost with handrail use for stroke survivors surpassed the reduction in energy cost due to external lateral stabilization in able-bodied individuals (7.5%), and even comes close to the relative contribution of balance control to the energy cost of walking in the most challenging situation in Chapter 3 (20.5%). Thus, these studies show that energy demand for balance control is indeed increased in the stroke population, and that balance control during ‘normal’
walking in stroke survivors inflicts a metabolic load comparable to that of challenging situations in able-bodied individuals.

**SOURCES OF THE ENERGY COST FOR BALANCE CONTROL**

Next to establishing the energy cost for balance control, we ventured to understand the origin of this cost. In the general introduction, we introduced potential sources of this energy cost, such as: a foot placement strategy, augmented muscle activation for a lateral ankle or hip strategy, or (co-)activation to increase joint stiffness. In this section we will review these mechanisms that can account for the energy cost for balance control, and where possible estimate the contribution of these strategies to the energy cost for balance control. As the lower limb amputees appeared to be hindered by the stabilization device, it does not seem valid to relate the gait changes in this experiment to balance control. Therefore this section will focus on the studies in able-bodied individuals and stroke survivors.

*The energy cost for foot placement control*

A good candidate as a source for the metabolic cost for balance control is the foot placement strategy. As explained in the general introduction foot placement is one of the primary balance control strategies through which gross control can be exerted. Balance control during walking is mostly needed in the medio-lateral direction. The foot placement strategy in this direction is apparent in step width, which is varied from step to step to control the margin of stability in the face of internal or external perturbations 45. In general, wider steps result in a larger margin of stability 46. However, a larger step width involves a larger metabolic cost for redirecting the center towards the opposite side, which increases with step width squared 18, 33. Because of this quadratic relation, increasing step width variability is also expected to increase the energy cost.

By facilitating balance control, a foot placement strategy is no longer crucial to ensure stability, and it could thus be expected that this will lead to the adoption of a step width which is (more) favorable with respect to energy cost. Accordingly, step width, step width variability and sideward pelvic motion decreased with increasing
stiffness of the stabilization device in able-bodied individuals, with an accompanying 7.5% reduction in energy cost (Chapter 2). In absolute values, mean step width reduced from 12.5 cm during normal walking at preferred speed, to on average 9.3 cm when walking with springs 2-4. Based on earlier work by Donelan et al. 18, such a decrease in mean step width, without taking into account changes in step width variability, would reduce the energy cost of walking by about 3% (based on a leg length of 0.9m). This shows that in able-bodied persons walking in situations with minimal challenge to balance control, a substantial part of the energy cost for balance control can be explained by the cost involved in step width control.

In the stroke patients, facilitation of balance control via handrail hold also caused a reduction in step width and step width variability consistent with a decrease in the margins of stability 161. With handrail support, step width reduced to values similar to those in able-bodied individuals during normal walking, from 17 cm to 13 cm on average. In a crude comparison using the data of Donelan et al again, this reduction is equivalent to a reduction in metabolic cost of ~4% of the energy cost of walking for able-bodied people 18. Taking into account the difference in energy cost of walking between stroke survivors and controls, this would constitute only ~1.7% of the energy cost of walking in the stroke survivors. Again, the decrease in step width variability in this study will add to this amount. Based on these values it can be estimated that in contrast to the able-bodied individuals, the contribution of step width control to the total reduction in energy cost (11%) was far less in stroke patients. This indicates that other balance control strategies, besides step width control, play a larger role in balance control and the related energy cost during walking in stroke survivors.

Another strategy to increase the margins of stability in the medio-lateral direction is to increase cadence (i.e. decreasing step or stride time) 46. When providing external stabilization to able-bodied participants, no changes in stride time or length were observed. In challenging situations we did observe a small (2.2%) decrease in stride time (Chapter 3), in line with an increase in margins of stability. Larger changes were observed in stroke survivors, in which the stride time increased by up to 16.1% when walking with handrail support relative to unsupported walking. These results
indicate that step frequency becomes more important for balance control in challenging situations and patients.

Changing step frequency also has metabolic consequences. Able-bodied people tend to walk at a step frequency-length combination close to the predicted resonant frequency of the legs. This frequency has been suggested to require minimal muscular activation and minimizes the metabolic cost of walking. It follows that deviation away from this resonant frequency results in an additional energy cost. Predicting the magnitude of the effect of a change in step frequency on the energy cost of walking in our studies is not easy, since changing step frequency when walking at fixed (treadmill) speed invariably leads to a change in step length, both of which can alter the metabolic cost of walking. However, empirical findings in able-bodied individuals show that a change in frequency from 1.37 steps s\(^{-1}\) to 1.17 steps s\(^{-1}\) which is close the one found in chapter 6 and at comparable walking speed (0.56 m s\(^{-1}\)) led to a substantial decrease in average energy cost of 0.74 J kg\(^{-1}\) m\(^{-1}\), albeit with large group standard deviations. If this can be compared to our stroke survivors, the change in step frequency would explain the majority of the total energy cost reduction due to handrail support.

Muscle activation for other balance control strategies

Looking at changes in the timing and position of foot placement as the sole explanation for altered metabolic balance control demands does not do justice to the complexity of balance control during walking. This is underscored by the discrepancy between the relatively minor changes in spatiotemporal step characteristics and the substantial increase in energy cost when walking under postural threat (Chapter 3). Moreover, in the challenging environment of chapter 3, step width even decreased while energy cost increased. The increase in the metabolic cost of walking when limiting the use of a step strategy also suggests that, although there is an energetic cost involved in the step strategy, this is a relatively inexpensive balance control strategy.

In the challenging balance control situations in Chapter 3 it was observed that increasing postural threat resulted in an increase in muscle activation amplitude and increased muscle co-activation of the main muscles of the lower limb in the two
highest threat conditions. Effects in the opposite direction were observed when facilitating balance control in stroke survivors, resulting in a global decrease in muscle activation, with decreased co-activation. The increased muscle activation of lower limb muscles in challenging situations, either due to external circumstances or pathology, can serve the foot placement strategy but also other balance control strategies, such as a lateral ankle or hip strategy, and co-activation of muscles to enhance joint stability. The additional muscle activation, required for these strategies results in an added metabolic cost. Unfortunately, with the current studies we are not able to disentangle muscle activation for these balance control strategies from the muscle activation related to step adjustments. Moreover with the current data no quantitative estimations on the metabolic equivalent of the increase muscle activation levels can be made.

**Conclusion**

In able-bodied people walking in situations with low challenge, step width control appears a major contributor to the energy cost for balance control. Although this strategy involves a metabolic cost, it is relatively inexpensive compared to other balance control strategies, as evidenced by the increase in cost when making this strategy unavailable. More challenging situations result in larger muscle activation, likely due to an increased reliance on other balance control strategies such as step frequency control, an ankle or hip strategy, or increased (co)activation of muscles. The increased muscle activation required for these strategies will add to the energy cost for balance control. Similar effects occur in stroke survivors in which ‘normal’ walking is already a challenge, requiring adaptations in gait which are in general similar to those of able-bodied people walking under challenging conditions. These gait changes can be viewed as adequate responses of the neuromuscular system to changing balance control demands, despite the expense of a higher metabolic cost. Disentangling the individual contribution of different balance control strategies to the energy cost for balance control, however remains a challenge for the future.
METHODOLOGICAL CONSIDERATIONS

Participant bias

To accurately determine the energy cost of walking it is imperative that the participant reaches steady-state oxygen consumption. This is usually reached after two minutes of walking, thus our walking trials lasted at least four and preferably five minutes. For many patients, especially those with a lower walking ability, such a prolonged walking activity is not possible. This introduced a bias in the selection of the lower limb amputees and stroke survivors towards those with a better walking ability. Since the walking ability of the stroke survivors considerably influenced the effect of cane use on the energy cost of walking, with larger benefits for more impaired patients, even larger effects are expected in those who were not eligible for this study. It would be interesting to see whether a valid and reliable estimation of the energy cost of walking can be attained based on shorter time periods, so that the energy cost for balance control can be measured in a larger cohort of patients with a larger variation in gait impairment.

Experimental manipulations

The explanation of the results regarding energy cost and gait changes presented above is built on the premise that the experimental manipulations target only balance control demands while leaving other mechanical demands of walking, such as the cost for body weight support, propulsion or leg swing, (largely) unaltered. Here, we will briefly review the potential and possible improvements of the different manipulations used in the studies presented in this thesis.

In Chapter 3 we intended to induce considerable challenge to balance control without requiring a substantial amount of energy to counteract mechanical perturbations. Therefore we only applied three perturbations in the final two minutes over which energy cost was calculated. Applying more, or even continuous perturbations, will likely increase the challenge, but will also require more external energy to be dissipated, directly leading to an elevated energy cost. We also avoided restricting path width to the extent that it would the projected path. The projection of
a path on the treadmill was used to vary the challenge to balance control without necessitate a change in the preferred gait pattern in order to walk within the path. Walking on a projected path requires subjects to attune to visual information, which potentially in itself alters the gait pattern. Including a condition in which path width is varied without applying external perturbations, provides the opportunity to differentiate between effects of postural threat and effects of attuning to visual information.

The external stabilization set-up used in Chapters 2 and 4 was based on previous research and appeared a promising and elegant method of stabilizing human gait in the medio-lateral direction. The underlying assumption was that the medio-lateral movement of the body during walking predominantly serves balance control, and that external stabilization by means of opposing and reversing lateral motion of the pelvis would facilitate balance control without having detrimental effects on walking. The work with the lower limb amputees, and pilot work with stroke survivors invalidated this assumption. Interestingly, others have recently been able to reduce the cost of walking in spinal cord patients by 10% with a similar set-up. This indicates that the problems with the set-up might be related to compensatory pelvic movements specific to certain patient populations.

Because of the difficulties with the external stabilization set-up in patients, we used a cane and handrail to facilitate balance control in Chapters 5 and 6. Canes or other walking aids are often prescribed to patients with balance control problems. However, a cane can also function as weight support and propulsive device, both of which could also facilitate increases in step length and step time. Since the exerted forces on the handrail during the handrail hold condition were small (largest forces were in the vertical direction and were on average 6.7% of body weight), we expect these effects to not play a major role.

In conclusion, manipulating balance control demands in isolation while leaving other task demands during walking unchanged has remained a challenge throughout this thesis. It can even be questioned whether it is fundamentally possible to fully disintegrate the cost for balance control from ‘other costs of walking’. Movements of the limbs and trunk can serve more than one purpose at a time, and facilitating or obstructing one task can inadvertently hamper the execution of other tasks. A
complete disintegration of the cost of walking into its parts may therefore not be feasible in ‘in vivo’ conditions. Therefore, some caution is advised when linking the observed changes in energy cost and the gait pattern to balance control.

**FUTURE STEPS**

The current thesis has identified the cost for balance control as a potential factor in the increased cost of walking in patient populations. However, from the above it is evident that properly establishing this cost, and identifying the sources of this cost has remained difficult. This opens up several avenues for future research.

First, a valuable step for the future is the refinement of the experimental manipulations to obtain a more reliable estimate of the cost for balance control. One example here is to adjust the external lateral stabilization set-up in such a way that it allows pelvic rotations, and perhaps even some medio-lateral motion. In pilot work with two stroke patients, we experimented with a set-up with an inner and outer frame, which diminished the restriction of these pelvic motions. Unfortunately, this did not appear to have the desired effect: the set-up still hindered participants. Recently, a similar set-up was developed by others, with a spherical gimbal allowing pelvic rotation around all three axes. This particular set-up was equipped with an admittance-controlled servomotor through which it is possible to provide balance support only when the pelvic pattern deviates from a reference pattern. The selection of a valid reference pattern representing ‘stable’ walking may still be difficult to define in patients with abnormal movement patterns, but it would be interesting to see the effects of such a set-up in a patient population.

Second, further disentangling the sources of the energy cost for balance control is necessary to understand where the energy cost for balance control originates from. Humans can and do employ multiple balance control strategies at the same time, making it difficult to estimate the individual contribution of these strategies to the energy cost for balance control. A promising field of research in this regard is that of musculoskeletal modeling of human walking. In the past decades there has been a rapid development of musculoskeletal models of normal and pathological gait. Work remains to be done to optimize musculoskeletal modeling such that the models can accurately predict (changes in) metabolic energy cost and efficiency. However in
the future, the use of forward dynamic simulations of human gait could allow for a better understanding of the sources of the energy cost of (pathological) gait in general, and the contribution of variance balance control strategies to this cost in particular.

Third, in this thesis we investigated the process of balance control but not the resulting gait stability. This is not without reason as stability is a complex construct that can be operationalized in many ways. Gait stability can be separated into different aspects, such as the size of a perturbation a person can accommodate (robustness), and the speed with which a perturbation can be counteracted (performance). Both of these aspects are the result of properties of the system (for instance joint range of motion or muscle strength), as well as the specific movement pattern. Even though balance control serves the purpose of ensuring gait stability, the effort for balance control is not one-to-one related to gait stability. For example, notwithstanding the increased metabolic effort for balance control, the results regarding the perturbation responses in Chapter 3 showed that participants needed more time to resolve a perturbation in the high threat condition, in other words, a decreased performance. Evaluating the relation between gait stability parameters and the metabolic effort for balance control was not an explicit aim of this thesis, but is an interesting area for future research.

Finally, with this thesis we aimed to contribute to an understanding of the elevated cost of pathological locomotion. Another way to address this issue is to take a longitudinal approach and evaluate changes in gait and gait economy over time. An evaluation of time related changes in the kinematics, kinetics, muscle activity and energetics of gait, either with a specific focus on balance control, or in a much broader sense, can pinpoint gait changes related to changes in energy cost. This can in turn help to design targeted experimental manipulations or interventions to improve our understanding of the energy cost of pathological gait.

**CLINICAL IMPLICATIONS**

The elevated energy cost of walking after a lower limb amputation or a stroke can be an important limiting factor for community ambulation, which deserves attention in the clinic. Currently, most research in this regard has focused on the effect of
technological advancement in prosthesis or orthoses development aimed at improving propulsion or weight bearing, with varying success. This thesis has provided evidence that an increased cost for balance control could be a contributing factor to the decreased economy in pathological gait, which should not be overlooked. To determine the energy cost for balance control in individual patients, a more accurate assessment tool needs to be developed, for instance through refinement of the external stabilization set-up.

Although designing a therapeutic intervention to reduce the energy cost for balance control was not an aim of this thesis, a clinical implication derived from this thesis is that training of balance control deserves a prominent role in gait rehabilitation, not only to reduce fall risk, but also to improve gait economy. Several interventions have been proven effective in improving balance after stroke, and effects of such interventions on the energy cost of walking should be investigated in the future. As an alternative to balance training, we have shown that providing a walking aid, such as a cane, can result in meaningful reductions in the energy cost of walking. Many patients and physicians are hesitant in prescribing or using a walking aid because of fear of detrimental effects on the gait pattern, or for cosmetic reasons. However, walking aids can provide a significant energetic benefit, next to an obvious safety/stability benefit, that may enable these people to enjoy participation in walking activities that would otherwise not be possible.